The HEELFOOT

DESIGN OF A PLANTARFLEXING PROSTHETIC FOOT

Freek te Riele

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PROEFSCHRIFT

ter verkrijging van
de graad van doctor aan de Universiteit Twente,
op gezag van de rector magnificus,
prof. dr. F.A. van Vught,
volgens besluit van het College voor Promoties
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door

Frederik Lambert Silvio te Riele

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te Dordrecht

Dit proefschrift is goedgekeurd door:

De promotor: Prof. ir. J.C. Cool

Voorwoord

Het ‘Highly Efficient Energetic Locomotion foot project’, dat wordt beschreven in dit proefschrift is in oktober 1998 van start gegaan en is gefinancierd door de Technologiestichting STW. Met veel plezier heb ik gedurende het project gewerkt aan de ontwikkeling van een nieuw soort prothesevoet en, tot mijn grote genoegen, heb ik ook nog mogen meedenken over het statisch balanceren van mechanismen. De hoofdmoot bestond echter uit de ontwikkeling van een prothesevoet, hetgeen ook tot uiting komt in dit proefschrift.

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Heerhugowaard, september 2003

Thesis Set-up

Originally the ‘Highly Efficient Energetic Locomotion foot’ project was funded for two researchers: One who would dedicate time to describe -and find applications for static balancing mechanisms in the biomechanical engineering field and one who was more dedicated to the design of a new prosthetic foot. Due to circumstances however the author of this thesis had the pleasure of doing both. This is why this thesis is divided in two sections: Section 1 focuses on the design of a new kind prosthetic foot. Section 2 discusses the design synthesis of several mechanisms that statically balance gravity acting on 1dof and 2dof inverted pendulums by means of springs.

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Abbreviations

3C-level	Level of comfort control and cosmetics
CF	Conventional Foot
COG	Center OF Gravity
dof	degree of freedom
ESF	Energy Storing Foot
FBD	Free Body Diagram
FF	Foot Flat
g-spring	Gastrocnemius-spring
GRF	Ground Reaction Force
HS	Heel Strike
HSW-model	Homo Sapiens Walking model
ICR	Instantaneous Center of Rotation
NME	Net Mechanical Energy
PTB-socket	Patellar Tendon Bearing-socket
SWS	Self-selected Walking Speed
TO	Toe Off
TT	transtibial (=through the lowerleg)
WC	Walking Cycle

SECTION 1:

Design of a Plantarflexing Prosthetic Foot

Abstract

Walking is a cyclic process during which gravitational, elastic and kinetic energies are exchanged from left to right and vice versa, via the trunk. The exchange of energy with its surroundings is minimal and the sum of all energies remains approximately constant. During this process the different body segments display cyclic oscillating movements that are alternately supported by both legs. If somewhere during stance a leg is too short, these oscillations are disturbed because they cannot be supported in the desired position. With a prosthetic walker this occurs mainly during late affected stance when the affected leg is in a backward rotated position and the prosthetic foot does not establish any plantarflexion (stretching of the foot). A new balance does set in, but at the cost of walking symmetry and efficiency. The prosthetic walker tries to compensate for -or reduce this asymmetry with excessive movements and loading of the remaining structures on both the sound and prosthetic side, resulting in increased energy consumption, early fatigue and decreased walking comfort. Furthermore, the continuous asymmetric and increased loading of remaining structures increases the risk on secondary complaints in the long run (such as lower back pain and early degenerative arthritis).

By elongating the lowerleg during the end of prosthetic stance by means of plantarflexion of the prosthetic foot, a transtibial-amputee can establish enough leg-length to sufficiently support the oscillatory movements of the body segments (in particular the trunk) during the end of affected stance. The typical power-burst that accompanies this plantarflexion action, allows for enough energy to be transferred from the affected to the sound side during the end of affected stance. This allows for a more natural progression and support of the trunk, resulting in a high valued prosthetic 3C (cosmetically sound, comfortable, and controllable) level during walking, and reducing the chance on long-term secondary complaints.

In this section of the thesis a prosthetic mechanism is presented that achieves leg-elongation during the end of the stance phase by means of plantarflexion of the foot. The energy needed for establishing the plantarflexion is borrowed from and given back to the total prosthetic-anatomic system in which human walking is considered to be a ballistic process. The proposed mechanism consists of a four bar linkage equipped with a flexible link or joint. The flexible link is chosen such that the prosthetic ankle joint has a rotational stiffness such that the gravity acting on the amputees' Center Of Gravity –rotating about the ankle joint, at a distance R - is approximately statically balanced. To further improve walking comfort and control, the prototype prosthesis also allows for a compliant coupling of the anatomic knee and the prosthetic ankle joint rotation to be established during walking. This is achieved by extending the prosthesis with an orthosis part.

Until now two prototypes have been preliminary tested. Prior to clinically testing, the first prototype's working was tested using a Homo Sapiens Walking model to acquire more insight in its' working during walking. The model consisted of a Homo Sapiens that was provided with braces underneath which prosthetic feet can be mounted. This enabled the Homo Sapiens

model to walk with prosthetic feet underneath his sound leg; on the left brace the prototype was mounted without shoe and on the right brace an Otto Bock single axis foot was mounted (with shoe). The HSW-model subsequently walked on a treadmill and the segment rotations of the prototype foot were recorded while the trial was video taped. These trials verified the hypothesized working of the prototype prosthesis (foot displayed plantarflexion and knee-ankle coupling increased walking comfort). More importantly however, these trials provided insight in the –possible- prototype prosthesis working during –and effects on walking prior to clinical trials.

A lighter version of the first prototype was tested with two transtibial amputees in a laboratory environment while biomechanical gait data was gathered. To allow for better assessment of the mechanisms working under walking conditions, these preliminary measurements were conducted without a shoe on the affected side.

The results of these preliminary tests again confirm the hypothesized working of the prosthetic foot: Plantarflexion (up to $\sim 18^\circ$) under loading was established and biomechanical analysis reveals a significantly increased affected ankle power-burst during the end of affected stance. Finally, an overall decrease in sound side Net Mechanical Energy generation and absorption - and peak joint torques suggests an overall decrease in sound side loading during weight acceptance.

In walking with the prototype both amputees felt a better push off and less resistance during the end of the affected stance compared to their current prosthesis. However, the short wearing time prohibited them to determine long-term 3C-level improvement. Finally, approximate center of gravity balancing proved to be a good guideline for determining the initial ankle joint stiffness (in combination with the proposed prototype foot).

When walking with the prototype foot in combination with the knee-ankle coupling, biomechanical data reveals little difference in joint kinetics and kinematics. Although it is seen that each amputee reacts differently on the knee-ankle coupling regarding the affected knee kinetics, both amputees felt that walking became less strenuous and more comfortable. The fact that an orthotic extension is required for the knee ankle coupling remains an issue -from a cosmetic and easy donning and doffing point of view- that needs to be tackled before the knee-ankle coupling will be worn outside a laboratory environment.

It can be concluded that the proposed prototype lowerleg prosthesis' mechanical working was as hypothesized during clinical trials. Also, the reactions of the subjects on walking with the prototype prosthesis were promising (for both with and without the knee-ankle coupling).

The new prosthesis is called the Highly Efficient Energetic Locomotion foot (HEELfoot) after the project name. A Dutch patent has been granted on the working of the HEELfoot and an international (PCT) application is pending.

Samenvatting

Lopen is een cyclisch proces waarin gravitatie, elastische en kinetische energie worden uitgewisseld van links naar rechts, via de romp. De uitwisseling van energie met de omgeving is minimaal en de som van de energieën blijft ongeveer constant. Gedurende dit proces worden de oscillerende bewegingen van de verschillende lichaamsdelen beurtelings ondersteund door beide benen. Als gedurende de standfase het standbeen te kort is, worden deze oscillaties verstoord omdat deze niet kunnen worden ondersteund in de juiste positie. Bij een protheseloper gebeurt dit vooral gedurende het laatste deel van de prothese standfase wanneer het prothesebeen schuin naar achteren staat en de prothesevoet geen plantairflexie (strekken van de voet) verkrijgt. Er stelt zich wel een nieuwe balans in, maar ten koste van loopsymmetrie -en efficiëntie. De protheseloper probeert deze asymmetrie te compenseren met extra bewegingen en (over) belasten van gezonde lichaamsdelen aan zowel de prothese als de gezonde kant, wat resulteert in een toename in energieverbruik, vervroegde vermoeidheid en een verminderd loopcomfort. Bovendien verhoogt de continue asymmetrische -en toegenomen belasting van de gezonde lichaamsdelen de kans op gezondheidsklachten op de lange duur (zoals lage rugklachten en gewrichtsslijtage).

Door het onderbeen gedurende het einde van de prothese standfase te verlengen door middel van plantairflexie van de prothesevoet, is een onderbeengeamputeerde in staat voldoende beenlengte te genereren om de oscillerend bewegende lichaamsdelen (met name de romp) op een juiste wijze te ondersteunen. De kenmerkende vermogengeneratie die gepaard gaat met deze plantairflecterende actie, maakt het weer mogelijk om voldoende energie van de prothese –naar de gezonde kant te transporteren gedurende het einde van de prothese standfase. Dit maakt een meer natuurlijke voortang en ondersteuning van de romp weer mogelijk, wat resulteert in een hoog 3C (cosmetisch, comfortabel en controle) niveau gedurende het lopen, en de kans op lange termijn gezondheidsklachten verkleint.

In dit deel van het proefschrift zal een prothese mechanisme worden gepresenteerd die door middel van plantairflexie van de voet beenverlenging bewerkstelligt aan het eind van de standfase. De voor deze plantairflexie benodigde energie wordt geleend van -en teruggegeven aan het gehele prothese-anatomische systeem waarin het menselijk lopen wordt beschouwd als een ballistisch proces. Het mechanisme bestaat uit een vier stangen mechanisme met een flexibel element. Het flexibele element is zo gekozen dat het prothese enkelscharnier een zodanige rotatiestijfheid krijgt dat de zwaartekracht die werkt op het zwaartepunt van de geamputeerde –roterend om het enkelscharnier, op een afstand R- bij benadering statisch wordt gebalanceerd. Om loopcomfort en controle nog verder te verbeteren, staat de prototypeprothese toe om de rotaties van de anatomische knie en de prothese enkel flexibel te koppelen. Dit wordt bewerkstelligd door het prothesedeel te verlengen met een orthesedeel dat afsteunt op het bovenbeen.

Er zijn tot nu toe twee prototypes voorlopig getest. Alvorens klinisch te testen, is het eerste prototype getest met behulp van een Homo Sapiens Loop model (HSL-model) om meer inzicht

te krijgen zijn gedrag gedurende het lopen. Het model bestond uit een Homo Sapiens die was uitgerust met stelten waaronder prothesevoeten kunnen worden bevestigd. Dit maakte het mogelijk voor een Homo Sapiens om te lopen met prothesevoeten onder zijn gezonder benen; onder de linkerstelt was het prototype bevestigd (zonder schoen) en onder de rechterstelt een Otto Bock enkel assige voet (met schoen). Vervolgens werden tijdens een loopsessie van het HSL-model op een loopband de verdraaiingen van de verschillende segmenten van de prototypevoet opgenomen terwijl de sessie op video werd vastgelegd. Deze sessies onderschreven de beoogde werking van de prototypeprothese (de voet bewerkstelligde plantairflexie en de knie-enkel koppeling verhoogde het loopcomfort). Belangrijker echter was dat deze sessies meer inzicht gaven in het –mogelijke- gedrag van de prothese gedurende –en het effect op het (pathologisch) lopen, voorafgaande aan de klinische sessies.

Een lichtere versie van het eerste prototype is getest met twee transtibiaal geamputeerden in een laboratorium omgeving terwijl biomechanische loopdata werd verzameld. Om beter te kunnen beoordelen hoe het prothese mechanisme zich gedraagt onder loopcondities, zijn deze voorlopige metingen gedaan zonder schoen aan de prothese kant.

De resultaten van deze voorlopige testen onderschrijven ook de beoogde werking van de prothesevoet: De voet plantairflecteerd (tot $\sim 18^\circ$) onder belasting en de biomechanische analyse laat een significante toename van de prothese-enkel vermogensgeneratie zien. Tot slot suggereert een globale afname van de netto gegenereerde –en geabsorbeerde mechanische energieën en gewrichtsmomenten, een globale afname van de belastingen aan de gezonde zijde gedurende gezonde gewichtsacceptatie.

Tijdens het lopen met de prothese voelden beide geamputeerden een betere afzet en minder weerstand aan het eind van de prothese standfase vergeleken bij hun huidige prothese. Door de korte draagtijd waren zij echter niet in staat aan te geven of dit een verbetering was van het 3C-niveau of niet. Tot slot bleek het bij benadering balanceren van het zwaartepunt een goede richtlijn voor het bepalen van de initiële stijfheid van het enkelscharnier (in combinatie met de prototypevoet).

Tijdens het lopen met de prototypevoet in combinatie met de knie-enkel koppeling laat de biomechanische data alleen kleine verschillen zien in gewricht kinematic en kinetiek. Alhoewel de knie kinetiek laat zien dat beide geamputeerden anders reageren op de knie-enkel koppeling, bemerken beide dat het lopen minder inspannend en meer comfortabel werd. Het feit dat een orthetische verlenging nodig is voor de knie-enkel koppeling, blijft een probleem – vanuit een cosmetisch –en gemakkelijk aan –en uitdoen standpunt- dat opgelost moet worden als de knie-enkel koppeling buiten een laboratorium omgeving moet worden gedragen.

Het kan worden geconcludeerd dat de mechanische werking van de voorgestelde onderbeenprothese was zoals verwacht gedurende de voorlopige klinische metingen. Verder waren de reacties van de geamputeerden veelbelovend (voor zowel met als zonder knie-enkel koppeling).

De nieuwe prothese is vernoemd naar de projectaanvraag titel en heet de ‘Highly Efficient Energetic Locomotion foot’ (HEELfoot). Een Nederlands octrooi is inmiddels toegekend en een internationale aanvraag is ingediend.

1 Introduction

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1 Introduction

After an amputation, the amputated limb is –most of the time- replaced by an artificial limb (a prosthesis). The function of a prosthesis is highly dependent on individual needs and abilities of a patient (e.g. cause and level of amputation, activity level, age, occupation etc.). It can vary from purely appearance purposes (e.g. breast prosthesis) to a functional necessity for a patient to regain independence in performing activities of daily living.

With a transtibial (=through the lower leg) amputation, the amputee will be fitted with a lowerleg prosthesis. In general the user will be able to establish –prosthetic- walking after a revalidation period. However, walking with current prostheses still causes discomfort and an early fatigue of the transtibial (TT) amputee. The extent of this discomfort/ fatigue of course depends on the medical indication on which the decision to amputate was based; a person who has undergone an amputation due to diabetes can be expected to have a more sensitive and ‘problematic to load’ residual limb than someone who had to be treated for cancer or after trauma. Nevertheless, the characteristics of the lowerleg-prosthesis directly influences the walking pattern, and since a user has to make do with the socket and prosthetic foot provided, ample thought has to be given to the conception of a prosthetic replacement of –part of the- the amputated limb. This chapter gives a general introduction in current lowerleg prosthetics.

Section 1.1 gives a survey on the causes (etiology) and types and distribution of amputations (epidemiology). Section 1.2 discusses current transtibial prosthetics. Section 1.3 gives an overview of the prosthetic design considerations. Section 1.4 gives the aim of the study and the outline of the thesis.

1.1 Etiology and epidemiology of amputations

Amputation of –part of- a limb has a severe impact on the life of a person. Thus only if no other solutions are present, the decision to amputate is taken. The causes on which the medical decisions to amputate are based, can roughly be divided into four groups ^{2, 14, 15, 22}:

- (70%) Ischemia (localized tissue anemia due to obstruction of the inflow of arterial blood) mostly due to diabetes or arteriosclerosis.
- (22%) Trauma
- (4%) Revision of congenital deformities
- (4%) Removing of tumor

Ischemia that can cause amputation are varied, but the most common ones are vascular disease (arteriosclerosis) and diabetes. Vascular disease limits the circulation to the extremities. Diabetes, which affects blood sugar, can decrease the body's ability to heal itself.

Trauma resulting in amputation is most frequently related to motor vehicle accidents and industrial accidents.

Congenital disorders or defective limbs present at birth are not amputations, but rather are a lack of part -or all of a limb. A person with a limb deficiency can usually be helped by use of a prosthetic device. Sometimes amputation of part of a deformed limb or some other type of surgery may be desirable before the application of an artificial limb.⁸

Tumors of the bone, called osteosarcoma, can sometimes only be treated by amputation of the limb.¹⁹

The level of amputation can be at different places on the lower extremity. Generally these places are divided into the following major levels of amputation (Fig.1-1):

- Different levels of amputation through the foot
- The transtibial amputation (=through the lowerleg)
- Knee disarticulation (=through the knee joint)
- Trans-femoral amputation (=through the upperleg)
- Hip disarticulation (=through the hip joint)

Determining the precise level of these amputations depends on skin viability, the proximity of tumor, or other -pathological, anatomical and surgical- criteria.^{6, 18, 22}

Figure 1-1: Levels and percentages of occurring amputations.^{2, 6, 14, 15, 20, 22}

Figure 1-1 shows that the majority of amputations are performed on a transtibial (TT) level than any other kind of amputation. This is because surgeons nowadays try to preserve the knee joint whenever practical and possible. This was not always the case since in the 1960's there were more transfemoral (=above knee) amputations because it was believed that it would improve the chance on wound healing.^{14, 22} The advantage of preserving the knee joint is that a knee function is retained because of the remaining flexor and extensor musculature (Fig.1-2).

Somewhat over 50% of the prosthetic users are between 25 and 65 years old. About 35% is older than 65 and approximately 9% is younger than 25.¹⁷

It can be concluded that amputations occur mostly in elderly people groups secondary to vascular disease associated with diabetes or arteriosclerosis⁹, complicating the transfer of weight bearing loads from prosthesis to residual limb.

Figure 1-2: With a transtibial amputation the knee function is preserved because knee flexor and extensor musculature action remains possible.⁴

1.2 Current transtibial prosthesis

Typically a lowerleg prosthesis is built up as shown in Fig.1-3a. It consists of a socket, a shank, a prosthetic foot and two pyramid adapters that connect the socket and the prosthetic foot to the shank. Typically the socket is of the Pattellar Tendon Bearing (PTB) kind. To compensate for the stump shrinkage, ensure/ improve a broad distribution of weight bearing loads on the residual limb (total contact), and prevent the skin from rubbing, a silicone gel liner and/ or several stump socks are generally worn between the PTB socket and the residual limb. The pyramid adapters allow for the socket and the prosthetic foot to be set at different angles relative to the shank. In that manner the prosthetic foot can be aligned relative to the residual limb.

With a conventional (bypass) prosthesis the weight bearing loads are supported on the upperleg and/ or the ischium (sitting bone) by means of a thigh cuff (orthosis), thereby relieving the residual limb (Fig.1-3b). Sometimes this is necessary because of a badly healing or very sensitive residual limb (neuroma), but also through arthritis in the knee joint. The thigh cuff is

connected to a (PTB) socket with hinges (at the anatomic knee joint) to allow for anatomic knee flexion and extension.

In this section the socket, liner and current prosthetic feet are shortly discussed.

Figure 1-3: Typical lowerleg prosthesis set-up. a) Lowerleg prosthesis with a Patellar Tendon Bearing socket (Copied from Northwestern University Prosthetics-Orthotics Center.¹⁶ b) Conventional lowerleg prosthesis with thigh cuff.⁷

1.2.1 The Socket and Liner

The main task of the socket is to efficiently distribute the substantial forces transmitted by the total prosthesis among the residual limb during standing and walking. A well-designed socket is carefully shaped to channel the heaviest pressures to the tissues best able to withstand them with the broadest pressure distribution possible (total contact principle). Thus, pressure and shear stresses are applied on pressure/ shear tolerant areas (e.g. the patellar tendon), thereby relieving sensitive areas such as the bony prominences (fibular head, tibial crest, distal end of tibia), blood vessels and nerves (Popliteal fossa). Figure 1-4 gives an indication of the areas of the stump that are generally considered sensitive and tolerant for pressure loading.^{1, 10, 11}

Considering the variety in sockets with different ‘main pressure distribution areas’, these areas are very case dependent in practice. Thus each socket is customized to achieve the most

comfortable pressure distribution on the residual limb although the fabrication process is always roughly the same.¹⁴

Generally the socket is used -or equipped with some sort of socket liner. There are many types of socket liners available nowadays (e.g. silicone liners, gel liners, air inflatable liners), each slightly different from the other. However, the main task of a socket liner (which is sometimes built in the socket) is to provide cushioning for the bony prominences and to ensure a broad pressure distribution (or total contact) from the residual limb to the prosthesis. Furthermore, the liner compensates for the day-to-day volumetric residual limb changes to ensure an everyday sufficient fitting of the socket.

Figure 1-4: a) Areas on stump tolerant to pressure. b) Areas on stump sensitive to pressure.¹¹

Due to the tight fit (vacuum) between liner and residual limb, it generally also function as an excellent suspension mechanism for the total prosthesis (By means of a pin at its distal end, which can be fixed to a ratchet on the socket). Other reported advantages of the tight fit are; reduction of pistoning of the residual limb relative to the socket, which increases the sense of control/ stability and sensorial feedback; reduction of the perceived weight of the total prosthesis; reduction of deformation and therefore potential irritation of the residual limbs' skin during walking. Finally, perspiration of the residual limb is reduced due to impregnation of anti-persperant lotion.^{3, 5, 12, 13, 21}

1.2.2 Current prosthetic feet

Although a correct socket design is vital in transmitting forces from the stump to prosthesis, the characteristic of the pylon and prosthetic foot combination is also important. After all, forces exerted by the stump are transmitted –via the socket- to the shank, the prosthetic foot and finally to the ground (and vice versa). Especially during walking this becomes undeniably clear as the prosthetic foot provides the dynamic support (or alignment) of the total prosthesis.

Nowadays many prosthetic feet are on the market. These prosthetic feet can be roughly divided into two categories of which an overview will be given in this section; so called Conventional Feet (CF) and Energy Storing Feet (ESF). It should be noted beforehand that the influence of these types of prosthetic feet on TT-amputee gait will be discussed more closely in the next chapter.

Conventional feet

One of the most common conventional prosthetic foot is the SACH foot (Fig.1-5). SACH is an acronym for "solid ankle, cushion heel," which refers to the compressible heel that functions as a shock absorber during heel strike. It is the simplest type of non-articulated foot with a more or less rigid keel, providing "pseudo" ankle action only by the rubber (available in different densities) heel that compresses under load during weight acceptance. The SACH foot is prescribed frequently because it is inexpensive and robust.

Figure 1-5: a) Otto Bock SACH foot. b) Kingsley SACH foot with covering.

The single-axis foot (one-joint) provides for ankle action that is controlled by one -or two rubber bumpers either of which can be changed to permit more or less motion resistance as needed (Fig.1-6). The main advantage of this foot when compared to the SACH foot is that it allows for a more rapid foot flat just after heel contact.

Figure 1-6: Single axis feet: a) Otto Bock single axis foot b) Kingsley single axis foot

A disadvantage is that the single-axis foot requires frequent maintenance because dirt can foul the bumper mechanisms. As is also the case for the SACH feet, this foot only allows for little – pseudo- ankle motion (dorsalflexion) during ambulation because of the more or less rigid keel.

Dynamic Response or Energy Storing Feet

Figure 1-7: Energy Storing feet: a) The Flex Foot (Ossur). b) The Modular III from Ossur (95% energy return). c) College Park Foot (Tru Step).

Basically the dynamic response feet and the flexible keel/ multi-axial feet have the same action, only dynamic response foot display a higher mechanical efficiency (less hysteresis). By flexing a keel or compressing a rubber bumper during heel-contact and rollover, energy is stored. Springing back during unloading, the intent is re-using this energy during the end of the stance phase to improve the push-off during late stance. That is why dynamic response feet are often called energy storing feet. The flexible keel or compliant ankle bumper(s) allow for dorsalflexion to be established on the affected side during walking, which seems to allow for a more continuous and natural progression of the trunk (See next chapter). The main disadvantages of these feet are the high costs and sometimes low durability. In recent years, there has been a proliferation of new designs for dynamic response feet of which some examples are given in Fig.1-7.

Reflection

In the previous sections a review has been given of current prosthetic feet. Many other prosthetic feet are on the market today. However, except from a mechanical efficiency point of view, all prosthetic feet are basically the same: All feet can be considered as levers that deform

(some more, some less) under loading and flex back during unloading. It is furthermore seen that none of the current prosthetic feet establish a plantarflexing rotation (stretching of the foot) about the –virtual- ankle joint during late affected stance.

1.3 Aim of the study and outline of the thesis

One of the main goals of lowerleg prosthesis designing is to channel weight bearing forces to the tissues that can best withstand them (but never intended for such weight bearing). In channeling these weight bearing loads from the ground to the residual limb, the socket and the prosthetic foot characteristics are both important: The socket forms the actual interface between prosthesis and residual limb, and therefore is vital in the prosthetic design; the prosthetic foot characteristic however determines the dynamic alignment of -or channeling of loads to the total prosthesis and therefore for a large part determines the amputees' perception of the total prosthesis in the performance of daily living tasks (such as walking).

Due to the intricate nature of the prosthetic designing process -in which possible improvements on current procedures or prosthetic devices from one field are always bound by the -technical-status of other fields-, probably the best way to come to technologically leading and innovative advancements is to revise the entire process using a holistic approach. However -although the distinct influence of the residual limb/ socket interface on the walking pattern is recognized- this thesis focuses on the development of a prosthetic foot in which re-establishing of the plantarflexion action during late affected stance is considered to be a necessity.

Although difficult to assess prior to clinical trials because of its' subjective and intricate nature, the aim of the new prosthetic foot is to increase the overall 3C level during TT-amputee walking. This 3C-level is divided into the level of:

- Control: The level in which the amputee can control his prosthesis (E.g. proprioceptive feedback is improved with increased stiffness of the socket).
- Comfort: The level of comfort depends on e.g. pain, energy expenditure but also on the sense of control during standing and walking (E.g. cushioning of bony prominences in socket, stiffer/ weaker ankle joint).
- Cosmetics: Cosmetics is a word that is synonymous to appearance of the prosthesis during walking or the performance of daily living tasks. Generally it is important to the amputee that it looks as inconspicuous as possible.

1.4 Outline of Section 1

Chapter 2 discusses current TT-amputee gait more closely and different strategies of TT-amputees to overcome the lack of plantarflexion action during the end of stance are discussed. Chapter 3 presents the design process of the prosthetic foot from concept to prototype. The first

section discusses typical anatomic –and typical current prosthetic foot characteristics more closely, and the possible advantages of elongating the affected lowerleg by means of plantarflexion of the prosthetic foot will be explained. In the course of the second section, a plantarflexing prosthetic foot concept is presented of which the working during the later half of the affected stance phase will be validated numerically using typical anatomic ankle-foot characteristics occurring during walking at self-selected walking speed as a reference. The third section discusses the conception -and the effects of restoring the knee-ankle coupling functionality during walking; in the final part of this section, the total prototype prosthesis is presented. Chapter 4 discusses the results of the prototype working using a Homo Sapiens Walking model and during clinical trials. Finally in chapter 5 some concluding remarks are given.

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2 **Unilateral Transtibial Amputee Gait**

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2 Unilateral Transtibial Amputee Gait

Numerous prosthetic devices (especially prosthetic feet) are on the market these days for TT-amputees, each device aiming to raise the 3C-level with slightly different characteristics. To acquire some insight in the working of these devices during walking, often the amputees' gait pattern is analyzed and compared with e.g. the gait pattern established with another prosthetic provision, or healthy gait characteristics. In that manner a researcher can assess whether or not a change in function of a prosthetic device has the intended effect on gait characteristics in practice. Of special interest to a researcher -and already numerous measured- can be the relative segment angular rotations, velocities and accelerations established about the different anatomic joints during walking, but also the net joint torques and powers resulting from muscular (co) contraction about an anatomic joint can be determined. To determine muscular activity, sometimes EMGs (electromyograms) are used, and finally metabolic energy expenditure is estimated from measuring oxygen consumption. Last but not least however, the amputee can give his or her perception of the effect of the prosthetic characteristics on the 3C-level during walking. This chapter discusses general trends in unilateral (one-sided) TT-amputee gait, occurring during walking at Self-selected Walking Speed (SWS), as described in existing literature.

Section 2.1 gives a short introduction to gait kinematics and kinetics analysis. For more elaborate information on this subject, the reader is referred to literature.^{4, 8-11, 25, 28, 29} Section 2.2 gives the terminology and conventions customary in gait analysis (human walking). Subsequently sections 2.3 and 2.4 discuss some general trends in unilateral TT-amputee gait characteristics. Finally a resume of the discussed sections is given in section 2.5.

2.1 Introduction

Kinematics

One of the most popular techniques currently used for capturing clinical gait is videography. Typically these systems track light reflective markers with cameras. Each camera has a light source (infrared LEDs surrounding the lens), which is reflected back by the markers. The video signal is scanned into a computer with a certain frequency. Using video analyzing software, the 2D marker positions from each camera can be reconstructed. Subsequently the data from different cameras (at least two) –detecting the same marker- are combined to reconstruct the 3D positions of the marker in the viewing/ working space.

In human motion study, the reflecting markers are placed on anatomic landmarks. By subsequently recording these marker positions with a subject standing upright, the anatomic joint positions and different segment properties can be estimated using anthropometrical data.

Through recording the marker positions during a walking trial, the different joint positions can be estimated in time, relative to a global coordinate system. This enables the rotations of different segments about the joints (angular velocities, accelerations, positions in space) to be determined relative to standing upright. The marker placement protocol used during this project is enclosed in the appendix together with a description of the procedures followed for the –kinematic- data capturing.

Other techniques for capturing motion include accelerometry where accelerometers are placed on rigid anatomic landmarks. From the measured accelerations, the velocities and displacements can be determined by integrating once or twice respectively (inverse kinematics). Opto-electronics is much like videography, only now the infrared LEDs are used as markers. The light is captured by cameras to subsequently determine the 3D position in the viewing space. In cinematography high-speed cameras are used to capture movements on film. In electrogoniometry, electric versions of our common plastic goniometers are used for determining joint angles. The simplest form of a one degree of freedom electro-goniometer consists of a potentiometer placed in the hinge joint of two bars. One bar is mounted on a segment proximal to an anatomic joint where the other is mounted distally such that the potentiometer lies approximately on the axis of rotation of the anatomic joint. Since the potentiometer gives a voltage output that depends on the angle between the two bars, the angle between the segments can be estimated (after calibration). For a more elaborate discussion on the different motion capture techniques, the reader is referred to literature.^{9, 25}

Kinetics

With the walking kinematics determined, and the masses and inertial properties of the different segments estimated, it is theoretically possible to determine joint forces and torques and estimate the Ground Reaction Force (*GRF*) and its center of pressure in the *single* stance phase (pure inverse kinetics). However, determining the joint kinetics with this method causes large errors.⁴ Besides that, additional information is –always- needed to determine the joint torques because the mechanism is underdetermined in the *double* support phases when both feet are in contact with the ground (closed loop problem). This is why it is customary to measure the *GRF* during gait trials simultaneously with capturing the kinematic data by letting the subject walk over force plates. With the aid of Free Body Diagrams (*FBD*) of each segment and Newtons' second law, the net joint torques, powers and reaction forces can then be determined.

Since this thesis focuses on the foot, Fig.3-3 shows the *FBD* of a sound right foot in late stance as an example. The figure visualizes the forces acting on and the kinematics of the foot segment. The ground reaction force $\bar{F}_{gr}$ and the ground reaction torque $\bar{M}_{gr}$ are known (measured with force plate). With the kinematics of the foot determined and the inertial properties estimated, the only unknowns are the net ankle torque vector $\bar{M}_{ankle}$ and the net

ankle reaction force vector $\bar{F}_{ankle}$, which can be solved by applying Newtons' second law. With the joint torques $\bar{M}_{ankle}$ determined and the joint angular velocities $\bar{\omega}$ known the joint powers can subsequently be determined. Note that the ground reaction torque vector $\bar{M}_{gr}$ represents the torque caused by the friction between the foot and the ground.

Figure 2-1: Forces acting on the foot segment with which the net ankle torque $\bar{M}_{ankle}$ and forces $\bar{F}_{ankle}$ are calculated. Note that the ground reaction force $\bar{F}_{gr}$ and torque $\bar{M}_{gr}$ act on the foot sole (copied from Vaughan²⁵). $\bar{\alpha}$, $\bar{\omega}$ and $\bar{a}$ are the angular acceleration about, the angular speed about, and the acceleration of the center of gravity of the foot segment respectively.

The joint torque and force data are limited by the fact that only resultant torques and forces can be estimated. Especially with co-contraction of musculature about a joint (e.g. agonist/antagonist action) it is difficult to assess the contribution of the different muscle groups to the estimated net joint torque. With that contribution not known, it is not possible to determine the really occurring forces acting on the ankle joint and although it is possible to estimate the muscle *activity* with electromyograms (EMG), it is not yet possible to accurately determine the force exerted by a single muscle group during walking. A description of the procedures followed for the –kinetic– data capturing during this project is enclosed in the Appendix.

2.2 Terminology and conventions

Figure 2-1a shows the three different anatomic reference planes with a subject standing in the anatomic position. A plane corresponding to a side view is called the sagittal plane. A plane corresponding to a front or a rear view is called the frontal plane. The plane corresponding to a top or a bottom view is called a transverse plane.

Figure 2-2: a) Anatomic planes (body in anatomical position) and main directions. b) Names of rotations about the anatomical joints with their axis of rotation in the sagittal and frontal plane. Rotations are noted with double arrow points.

Common directional terms are: In front of = *anterior* to. Behind or backside = *posterior* to -or side. The outside of a leg = the *lateral* side. The inside of a leg = the *medial* side. Above = *superior*. Below = *inferior*. With respect to arms and legs sometimes the terms *proximal* and *distal* are used (close to -and further away from the trunk respectively), instead of superior and inferior.

Segment rotations -relative to their anatomic position- in the sagittal plane are referred to as either flexion or extension rotations. Figure 2-2b shows the convention for joint rotations in the plane of progression relative to standing upright: Hip flexion, knee extension and ankle dorsal flexion are considered positive rotations.

Figure 2-3: Different phases and definitions of a walking cycle (copied from Inman⁸).

Figure 2-3b shows the different phases occurring during a complete Walking Cycle (WC). The duration of a WC or stride is defined from Heel Strike to Heel Strike (HS) of the *same* leg. The figure shows that the stance phase of the right leg starts with right HS and ends with right Toe

Off (TO), successively the right swing phase is initiated which ends with a right HS, completing (or starting) the WC.

During a *double support phase* both feet are in contact with the ground. The right double support phase is the time between left HS until right TO. It is seen that the right stance phase starts with a left double support phase. With left TO, the left swing phase and right single support phase are initiated. With left HS, the right double support and left stance phase start, initiating the right swing phase and left single support phase. With the following right HS, a WC is completed.

The *stride length* is the horizontal distance covered along the plane of progression during one WC; it is the distance covered from HS to HS of the *same* foot (Fig.2-3a). The stride length is equal to the sum of the right and left step lengths.

The *step length* is the horizontal distance covered along the plane of progression during one step; A left step length is measured from the contact position of the right heel to the contact position of the subsequent left heel when they are in contact with the ground (Fig.2-3a).

2.3 Transtibial amputee gait

During this section some of the general trends of unilateral TT-amputee gait with current prosthetic feet will be reviewed for walking at a Self-selected Walking Speed (SWS). First the temporal trends en kinematics will be discussed, followed by the kinetics and energy expenditure of TT-amputee walking. Finally a reflection on the whole section is given.

2.3.1 Temporal trends and kinematics

When discussing temporal trends, things as SWS, cadence, stride length, step length, stance time, swing time etc. are discussed. These temporal trends are often used for comparing gait and the study of gait symmetry.

Self-selected Walking Speed and cadence

The SWS of TT-amputees is generally lower than normal.^{1, 2, 7, 19, 26} Table 2-1 shows that the etiology of the TT amputation influences the SWS and cadence⁷ as the SWS of a TT-amputee with underlying vascular disease is lower than for a TT-amputee secondary to trauma. The SWS of traumatic amputees is 80% of the control group whereas the vascular amputation group is a mere 53% of the control group. Furthermore it is seen that TT-amputee cadence also decreases relative to the control group.⁷ Other researchers reported similar results^{1, 2, 19, 26} whereas Sanderson²¹ reported no significant difference in cadence when comparing traumatic TT-amputees with his control group. This however is probably because the control group was forced to walk at 1.2 m/s. Although this walking speed is the approximate SWS of traumatic TT-amputees^{7, 26}, the SWS of normals is reported to be approximately 1.37m/s for men and

1.23m/s for women.^{3, 26} This most likely influences the cadence with which a subject walks. Finally it is seen that the stride length of a TT-amputee is also shorter than normal, where the vascular amputee has a shorter stride length than the traumatic amputee.

<i>Measurement</i>	<i>Group</i>		
	<i>Vascular</i>	<i>Traumatic</i>	<i>Normals</i>
<i>Velocity</i> (m/min)	44.7±3.2	67.2±4.8	83.5±3.2
<i>Cadence</i> (steps/min)	83.4±4.0	92.3±3.4	111.5±2.3
<i>Stride length</i> (m)	1.1±0.1	1.4±0.1	1.5±0.1
<i>Prosthetic limb single support stance</i> (%WC)	33.4±1.7	36.8±2.5	-
<i>Sound limb single support stance</i> (%WC)	34.7±1.6	38.8±2.3	38.0±1.0
<i>Prosthetic limb stance</i> (%WC)*		61.8±0.9	-
<i>Sound limb stance</i> (%WC)*		65.3±2.6	65.7±1.2
<i>Prosthetic limb swing</i> (%WC)*		38.2±0.9	-
<i>Sound limb swing</i> (%WC)*		34.7±2.6	34.3±1.2

Table 2-1: SWS, cadence, stride length, and prosthetic and sound limb stance times for the motion analysis of vascular (n=8), traumatic (n=8), and normals (n=8) (Huang et al⁷). * Sanderson²¹.

Stride-length, step-length and stance time

A healthy subjects' gait is generally symmetric. This means that stance time and step-lengths of both limbs are equal. Generally TT-amputees display a decreased affected stance time relative to the sound side, as table 2-1 also shows.^{2, 7, 13, 16, 19, 21} The decrease in the stance time on the affected side is accompanied with an increase in swing time.^{2, 13, 21} Furthermore it is seen that the stride length of unilateral TT-amputees decreases. Affected double support time is also reported to be shorter than sound double support time.^{2, 13, 16, 19}

With the step-length of a limb as defined in section 2.2, it is reported that the affected step-length of TT-amputees is slightly longer compared with the sound side.^{2, Robinson} This is common pathological gait. A cause for this may be a lack of trust of the affected side for weight bearing and to relief residual limb loading.^{2, 13} This will be further discussed in the kinematics section.

Heel contact period and Foot Flat

Powers reports a significantly longer heel only contact period on the prosthetic side (20%) when compared to normals when walking with a Seattle light foot. Since the Seattle light foot has a solid ankle, this is understandable. However Perry¹⁶ also reported some lateness in

achieving Foot Flat for subjects walking with a Seattle light, a single axis, and a flex foot. The prolonged heel only support results in an unstable base of support for persons with TT-amputations, resulting in a tendency of the TT-amputee to prolong the sound double support time.^{13, 16, 19}

Joint Kinematics

Figure 2-4 shows the –averaged- anatomic joint rotations relative to standing upright of 19 healthy subjects walking at SWS.^{28, 29} This section describes general kinematic trends of unilateral TT-amputee walking at SWS with these data as reference.

Figure 2-4: Inter-subject joint angles average for walking at SWS, projected on the sagittal plane.^{28, 29}

In comparing joint kinematics of the affected side of TT-amputees with their non-affected side and non-amputees, all studies reveal the lack of plantarflexion on the affected side during the end of the stance phase. This is because all current prosthetic feet can be considered as leaf spring like levers that deform (some more, some less) under loading and flex back during unloading. Therefore they are not able to establish a plantarflexion rotation about the –most of the times virtual- prosthetic ankle joint relative to standing upright.

Considering the trends in the joint angles of the hip and knee on the affected side generally more subtle differences are found. Sanderson reports that affected hip and knee angles are more in extension (in the sense that the upper -and lowerleg remain more vertically aligned) on the affected side when compared to the sound limb (and normals), particularly during the first two thirds of the stance phase. Other authors reported similar results.^{16, 19} The sound limb hip and knee angles are also slightly more in extension during the end of the stance phase when compared to normal. This is probably to increase the prosthetic step length and sound limb stance time because of the prolonged unstable Heel Only and to push the trunk over the slightly more extended affected limb (like pole vaulting).^{2, 21} Bateni² also reports increased hip flexion on both the affected and unaffected side during early stance, caused by the tendency of a TT-

amputee to lean forward to bring the center of gravity over the affected foot by moving the trunk forward in stead of ‘pulling’ or ‘pushing’ the trunk over the stance leg (knee flexion -and hip extension torque), thereby -most likely- aiming to reduce residual limb loading and to compensate for the lack of plantarflexing action during the end of affected stance.^{2, 21, 23, 27} Furthermore his subjects displayed higher peak knee flexion on the unaffected side during early stance, as was also the case for Sanderson²¹.

2.3.2 Kinetics

Contrary to the more subtle differences in kinematics (particularly of the hip and knee), some distinct differences are revealed when comparing TT-amputee gait -with normal gait kinetics. In this section the typical kinetic trends occurring during TT-amputee walking at SWS are compared with typical healthy walking at SWS.

Figure 2-6 shows the averaged kinetic data of 19 healthy subjects for walking at SWS. The measured Ground Reaction Force (GRF, Fig.2-6a), the derived net joint torques (Fig.2-6b), and the joint powers (Fig.2-6c) occurring during walking in the plane of progression are displayed. All kinetic data have been normalized to bodyweight.^{28, 29}

Figure 2-5: Convention for joint torques in the plane of progression (the sagittal plane). All clockwise acting joint torques, as calculated at the proximal end of each segment, are considered positive.²⁹ M_a , M_k and M_h are the net occurring ankle, knee and hip torque respectively.

Figure 2-5 shows the standard convention for the joints torques in the plane of progression. All clockwise acting joint torques, as calculated at the proximal end of each segment are considered positive and counterclockwise torques negative.²⁹ Thus knee flexion, hip extension, and an ankle plantarflexion torques are considered positive (Also see Fig.2-2).

A joint torque is the net torque resulting from (co) contraction of musculature about that joint to prevent the extremity from collapsing. Thus a knee extension torque will cause the knee to extend if an external knee flexion torque is not present (or is smaller in magnitude).

Ground Reaction Forces

In comparing the TT-amputee GRFs with the normal GRFs (Fig.2-6a) occurring during walking at SWS, it is generally seen that the –second- peak of vertical, but also both peak forward and backward GRFs displayed by TT-amputees are lower than normal on the affected side.²¹ Furthermore an increase of the first peak GRF is frequently reported for the sound limb when compared to the affected limb (during sound weight acceptance). An overall increase of vertical loading on the sound limb when compared to the affected side and to normals is even reported; despite of a TT-amputees' reduced walking velocity.^{1, 21, 22}

Net joint torques

For normal walking the ankle joint musculature exerts a relative small dorsiflexion torque during weight acceptance (to establish a quick and controlled foot flat; Fig.2-6b). At approximately 8%WC a plantarflexing ankle torque starts building while establishing a dorsiflexing rotation of the foot. After reaching its peak at approximately 48%WC the plantarflexion torque rapidly decreases until TO is established at approximately 62%WC (while establishing a rapid plantarflexion rotation of the foot).

Considering the affected ankle joint, generally lower than normal peak plantarflexor torques are measured during late stance (~2/3 of normal) whereas the sound limb shows a larger than normal plantarflexor torque. Furthermore, the dorsiflexor torque is both longer in duration and larger in magnitude on the affected side than is the case for normals, most likely caused by the prolonged heel only period.^{2, 21, 27} Sanderson²¹ however also reports a dorsiflexor torque that is both longer in duration and larger in amplitude on the sound side than is the case for normals.

Figure 2-6b also shows that the knee musculature for normals exerts a short flexion torque just after HS, which rapidly changes into an extension torque to prevent a collapse of the lowerleg as the knee flexes. The maximum knee extension torque is achieved at 14% WC, when the knee flexion angle is also at a maximum (during early stance). Subsequently the knee extension torque drops as the knee flexion angle decreases. At 44% WC, the knee flexion torque is maximal (with the knee in maximal extension). After that the flexion torque decreases, achieving a relative small extension torque prior to TO, to initiate swing. Finally this extension torque drops and becomes flexion in direction from 80-100% WC, to decelerate the lowerleg prior to HS.

In TT-amputee walking, the affected knee torque is generally reported to be lower or even near zero during the first half of affected stance. During the second half of affected stance -and swing however, the affected knee torque lies well within the normal range. Furthermore the affected knee torque is primarily flexor in direction.^{19, 21, 27} Sanderson²¹ reports a higher sound

knee extensor torque during late -and early stance (although the latter did not reach statistical significance) when compared to the affected limb, which concurs with his reported increase of both sound limb vertical and backward forces. Bateni² also reports higher than normal peak sound knee extension torque during early and late stance for all his subjects.

Finally, Fig.2-6b shows that the musculature acting about the hip joint exerts a net extension torque during HS that gradually changes into a flexion torque as the walking cycle progresses. A maximum in hip flexion torque is achieved at approximately 50% WC. After that the flexion torque decreases as TO is established and the upper -and lowerleg are accelerated into swing. From approximately 80-100% WC, the hip musculature exerts a net extension torque to decelerate the upper -and lowerleg prior to secondary HS.

Bateni² reports that the affected hip torque varied greatly among subjects, some showed greater than mean normal while others showed close to normal hip extensor action (considering the lower SWS this means that hip extensor torque is always higher relative to normal SWS walking). The sound side showed higher than normal mean peak extensor torque during early stance as well as higher than normal flexion torque during late stance and early swing.² These peaks approximately occur concurrently with the reported peaks in sound knee extensor torque. Suggesting extra energy is generated during early and late sound stance to compensate for the lack of plantarflexion action on, and to relief the affected side. Winter²⁷ reports highly variable inter-subject hip torque on the affected side and does not talk about the sound side (Affected hip torque varied from mainly extensor to flexor, to approximately normal). Sandersons' data²¹ also suggest an increased sound hip extensor torque compared to the affected side during early stance. His subjects however walked with Flex Feet, which are reported to result in higher peak values in the affected hip and ankle torques when compared to other prosthetic feet.²³ Finally, Powers¹⁹ reports increased affected hip extension torques during early affected stance because of 'a forward trunk-lean to augment forward progression over the relatively solid ankle'.

Net joint Powers

Figure 2-6c shows the net mechanical joint powers generated and absorbed about the anatomic joints for normals walking at SWS.^{28, 29} The convention of the mechanical joint power P is such that it is positive when the joint torque M and the joint angular velocity ω act in the same direction; the joint is said to generate energy. When the joint torque M and the angular velocity ω act in opposite directions, the joint is said to absorb energy and is considered negative.

With normals, it is seen that the mechanical energy generated about the ankle joint is about three times the energy that is absorbed; therefore the ankle is considered the main energy generator during a walking cycle. The hip generates about two times more energy than it absorbs during walking and can therefore also be considered as an energy generator. The knee however generates about 30% of the energy absorbed and can therefore be seen as the main energy absorber during walking. Inman⁸ reported similar results in his book.

Figure 2-6: Inter-subject data of 19 healthy subjects walking at SWS [Winter]: a) Average vertical (F_y) and forward/ backward (F_x) Ground Reaction Forces during stance. b) Average sagittal joint torques. c) Average net joint powers generated and absorbed.

When considering the net joint powers generated or absorbed of TT-amputees walking at SWS, the most striking difference with normals is that the affected ankle joint is mainly an energy absorber, almost totally lacking the typical power-burst during late stance that is so typical for the anatomic ankle.^{2, 5, 19, 27} This is as expected, since a prosthetic foot is essentially a leaf spring with –sometimes- excellent damping qualities and therefore is unable to generate energy. However, the sound ankle is reported to generate more energy than normal during late sound stance.^{2, 21} This is to prolong the sound side contact with the ground while augmenting forward progression of the trunk.

Generally, the affected knee displays near zero or significantly reduced power absorption/generation from early to late stance. From late stance (~45% WC), the affected knee displays normal^{5, 19, 27} or even increased² power absorption as the knee flexes while exerting an extension torque. The sound knee is reported to absorb more energy than normal during late stance.^{2, 21} Bateni² also reports increased energy absorption of the sound knee relative to the affected side during late swing. This is as expected since the prosthetic leg is lighter than the unaffected leg and that power absorption phase is to decelerate the lowerleg prior to HS (knee extension rotation while exerting a flexor torque).

Considering the hip joint, the affected side shows a stronger than mean normal power generation during the first half of the stance phase. This is –reportedly- to propel the body forward from the rear^{2, 5, 27} and to compensate for the lack of plantarflexion action on the affected side. However, it is also likely that this is a result from the slightly forward bending trunk position to augment its' forward progression and relief the affected knee. Bateni² also reports a higher hip power generation burst on the sound side during late stance, which is to elongate the sound limb stance and to compensate –partly- for the loss off power generation about the ankle; Sanderson²¹ reports similar results.

2.3.3 Energy expenditure

The energies/ powers discussed in the previous paragraph are distinctly mechanical energies, which are not to be confused with the metabolic energy consumed by the musculature to achieve the net mechanical powers about these joints.

The –metabolic- energy expenditure (cal) needed to establish a WC is generally estimated from the volume of O₂ consumption and CO₂ production. The rate of O₂ consumption is given in milliliters O₂ per kg bodyweight per minute (ml/kg per min). The physical work done in walking is derived from the rate of O₂ consumption in milliliters per kg per meter traveled (ml/kg per min).

Table 2-2 shows the O₂ uptake per meter and minute of normals, traumatic -and vascular TT-amputees walking at SWS. It is noteworthy that minimal O₂ uptake per meter, is related to walking at SWS by numerous investigators.^{20, 26} The data presented in the table suggests that

the O_2 uptake per meter decreases with increasing age. Furthermore TT-amputees display an increased O_2 uptake per meter than normals, the amount of O_2 uptake depending on etiology as vascular TT amputees consume more O_2 than traumatic TT amputees. However, the O_2 uptake per minute is about normal for the traumatic amputees whereas for dysvascular amputees the O_2 uptake per minute is even lower. This suggests that walking speed is adjusted such that – residual- muscle activity requires an approximately normal O_2 uptake per minute. Other investigators have reported similar results.^{1, 7, 15, 26}

<i>Measurement</i>	<i>Group</i>			
	<i>Normal adult</i>	<i>Normal children</i>	<i>Traumatic TT amputee</i>	<i>Vascular TT amputee</i>
<i>O_2 rate</i> (ml/kg per min)	12.1	15.3	12.4	9.4
<i>O_2 rate</i> (ml/kg per m)	0.15	0.22	0.18	0.21
<i>SWS</i> (m/min)	80	70	71	45

Table 2-2: Energy expenditure of normals, traumatic, -and vascular unilateral TT-amputees walking at SWS.²⁶

The increased O_2 uptake per meter can probably be contributed to the increased muscular activity that is measured on the affected side. EMG studies of the affected side of TT-amputees walking at SWS reveal above normal hip extensor activity (Hamstrings, Gluteus maximus), which are both longer in duration and higher in intensity than normal during with a peak occurring during early stance (weight acceptance).^{5, 19, 27} This is in accordance with the increased hip extension torque due to the forward trunk lean of the amputee.^{2, 19} Furthermore an above normal muscle activity of the quadriceps musculature on the affected side is also measured. Since hamstrings and quadriceps generate opposite torques, co-contraction of these muscle groups can be a major contributor to the increased metabolic energy expenditure while a relatively low net affected knee joint torque and power are measured.^{5, 7, 19, 21, 27} The main reason for this increased co-contraction is to stabilize the affected knee, especially during early stance (with the prolonged unstable Heel Only period).^{16, 19} Finally the overall increased sound side loading that is revealed during both early and late stance will also increase metabolic energy expenditure.

2.4 Transtibial amputee gait and energy storing feet

2.4.1 Temporal trends and stride characteristics

The majority of all studies reveals non-statistically significant or minor differences in temporal and stride characteristics although trends towards increased SWS and stride length when compared to conventional feet can be detected.⁶ Energy Storing Feet (ESF) display a prolonged mid-stance support time (delayed Heel Off), which is more similar to the sound side due to the flexible keel -or ankle joint. Affected limb stance time remains shorter than sound limb to shift weight from the affected limb to the sound limb more quickly. Also the sound side double support phase (from sound side HS to affected side TO) remains shorter than that of the affected side.¹³ Several investigators report a decreased sound side step-length due to the increased dorsalflexion about the -virtual- ankle joint^{1, 7}, other investigators report increased sound limb step-length^{13, 6}, and yet again Torburn²³ reports no difference. Although this seems contradictory, all results are feasible: The flexible keel or ankle allow for a smoother transition between rolling about the heel lever and rolling about the -tip of the- forefoot lever by enabling a controlled (compliant) dorsalflexion about the -virtual- ankle joint while decelerating the center of gravity, when compared to Conventional Feet (CF) with a rigid ankle joint. This can cause an increased sense of late stance stability and increased walking comfort, which shows in the willingness of the amputee to prolong the affected single support stance time.^{13, 14} Hence sound side step-length can be increased through dorsalflexion about the ankle joint. However, this dorsalflexion actually shortens the lowerleg under loading when compared to CF, meaning that TT-amputees fitted with CF can in fact establish longer sound side step-lengths provided they -are able to- make use of this increased lowerleg length. Finally, Macfarlane¹³ reports a decrease in vertical trunk displacement when walking with ESF compared to CF.

2.4.2 Kinetics

Ground Reaction Forces

Regarding the peaks in occurring Ground Reaction Forces, ESF seem to reduce sound limb weight acceptance force and an increase in the affected side propulsive force is frequently reported.^{6, 22, 23} This is most likely related to the smoother and more natural progression of the center of gravity that a flexible keel enables during ambulation (less vertical movement in combination with dorsalflexion about the ankle joint while the center of gravity's decelerates). Barth¹, on the other hand reported significantly higher sound limb weight acceptance forces for

subjects wearing the Carbon Copy II and the Quantum foot. These feet however allow for less dorsalflexion as does the Flex Foot, which was used in the other studies.

Net joint torques and powers

The main differences between CF and ESF lie in the foot characteristics themselves as increased –virtual- ankle dorsalflexion motion (close to normal) and peak affected ankle plantarflexor torques (close to normal) during walking with ESF feet are reported. Apparently, the amputee is able to establish higher peaks in affected ankle joint torques if the prosthetic foot allows for dorsalflexion about the ankle while a(n) (internal) plantarflexing torque gradually builds.^{6, 21, 23} ESF also display an increased affected ankle power-burst during late affected stance.^{2, 5} Note however that this power-burst remains substantially lower than normal. Generally these ESF characteristics only have minor effects on other joint kinematics and kinetics.

Considering knee kinetics, a lot of conflicting results are reported, which vary from mainly flexor to mainly extensor action of the affected knee during stance. These results acknowledge the subjective nature of the prosthetic users (group). Note however that the affected knee torque remains minimal throughout the first half of affected stance^{21, 23} (but especially just after weight acceptance). Compared to CF, ESF display little (non-significant) differences with respect to affected knee rotations.^{21, 23} However, Huang⁷ and Snyder²² do report increased sound knee flexion during early sound stance.

Regarding the effect of ESF on hip torques, Hafner⁶ reports few differences with conventional feet, but also reports several studies revealed increased hip extensor torques during weight acceptance [e.g. Torburn²³]. Considering hip rotations, Torburn²³ reports no significant differences on the affected side. Huang⁷ reports increased hip flexion rotation of the sound side compared to the affected side.

2.4.3 Energy expenditure

The improved mechanical energy efficiency of ESF over CF is frequently assumed to decrease the metabolic energy expenditure during TT-amputee gait. Also, ESF are generally lighter than CF which causes a proximal shift of the center of mass of the total prosthesis, which is beneficiary to prosthetic gait efficiency.¹² However, energy expenditure wise no significant changes are reported between conventional feet and energy storing feet during walking at SWS. A decrease in metabolic energy consumption is measured only for fast paced walking or higher level activities.^{1, 6, 15, 23, 26}

This is most likely because the ESF characteristics have little effect on knee and hip kinematics and kinetics for walking at SWS, suggesting the same excessive straining of remaining structures as for CF.

The relative decrease in metabolic energy consumption during fast paced walking is therefore more likely the result of the increased ankle range of motion offered by ESF (allowing for a more continuous progression of the COG during ambulation) than the increased mechanical efficiency.

2.4.4 Amputees' perception

Although there are little significant biomechanical data available that support the subscription of ESF over CF, the prosthetic user most of the time prefers an ESF over a CF. The ESF is reported to feel lighter, livelier, reduce early fatigue, and increase SWS. The main reason for that is the broader range of ankle motion of the foot due to the -energy storing and releasing-flexible keel -or ankle joint- action during ambulation, which seems to increase walking comfort and late stance stability and control on the affected side on uneven terrain, especially during higher level activities.^{6, 14, 15, 23, 24} In assessing the subjective ratings of TT-amputees regarding to ESF and CF, Postema¹⁷ concluded that no clear preference existed for either the ESF or the CF. Reported negative comments however include complaints of late stance instability and poor knee control, the need to relearn walking patterns and the sensation of falling off the front of the foot during TO.²⁴ In these cases the extra range of motion is that the energy storing foot supplies, is obviously the cause of the decreased sense of stability. Torburn²³ draws similar conclusions when at the end of his study, all of his subjects preferred an ESF feet over conventional, but none chose the Flex Foot with its' substantially broader range of ankle motion compared to the other feet (ESF and CF).

2.5 Resume

The previous sections on kinematics and kinetics described -possible- strategies that unilateral TT-amputees use to compensate for the lack of affected plantarflexing action during both the sound and the affected double support phases. It is reported that from early -until approximately two thirds of affected stance the affected knee powers and torques are minimal while an increased EMG activity of the affected hamstring and quadriceps is measured. During late stance however, affected knee powers and torques lie well within the normal range (Bateni² even reports an increase in affected knee extension torques during early affected stance). Furthermore kinematic study reveals a shorter affected stance phase, sound step-length, and the affected double support phase (from sound side HS to affected side TO). These typical gait characteristics mainly result from the lack of plantarflexion of the prosthetic foot during weight acceptance, but especially during the end of the affected stance phase.

During sound double support, the amputee reacts to a the prolonged -unstable- heel only period by prolonging his sound side push off action during the end of the sound limb stance

phase (prolonged sound hip, knee and ankle extension rotation compared to normal in combination with increased sound knee extensor, hip flexor and ankle plantarflexor torque). Simultaneously the center of gravity is pushed forward and upward, over the affected limb. To augment the progression of the center of gravity, the amputee also maintains a forward trunk-lean (increased affected hip flexion rotations in combination with increased affected hip extensor torques and powers). Since affected knee torques and powers are reported to be minimal during early affected stance while the EMG activity of the affected hamstring and quadriceps increased, it is likely that these actions are to relief the affected knee and the residual limb while stabilizing the knee through massive co-contraction of musculature acting about the affected knee joint.^{16, 19}

During the affected double support, the amputee compensates for the lack of affected side energy generation and leg-length by decreasing the sound step-length. Also due to lack of the 'pole vaulting like' plantarflexion action of the prosthetic foot, the amputee again maintains a forward trunk-lean to augment the progression of the center of gravity in combination with a pulling kind of action during sound weight-acceptance (increased sound hip -and knee flexion rotations in combination with increased sound hip -and knee extension torques). Since the affected knee characteristics during the last third to one half of affected stance are reported to lie well within the 'normal' range, these actions are probably not so much to relief the affected knee and residual limb as to compensate for the loss of affected leg-length -and energy generation.

In short, clinical trials with TT-amputees suggest the amputee compensates for the lack of plantarflexion action of the prosthetic foot during the double support phases, by extra (excessively) straining remaining structures, which results in an overall increase in sound limb loading and an asymmetric support of the trunk. The extra straining of the remaining structures probably causes the increased energy consumption of TT-amputees compared to normal (while SWS decreases).

When comparing gait of TT-amputees wearing CF with the same TT-amputees wearing ESF (for walking at SWS), ESF display increased affected ankle dorsalflexion -and plantarflexion torque during ambulation (because of the flexible ankle joint or keel). Other than that, literature shows only minor (mostly non-significant) and conflicting differences. Nevertheless, on the whole there seems to be a preference for ESF over CF by a large group of TT-amputees, especially in the performance of higher-level tasks (e.g. fast -or uphill walking). This is most likely because their broader range of ankle motion allows for a more continuous progression of the trunk (but also -to a lesser extent- because of their relative light weight).

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2.7 Appendix: Marker placement for VICON measurement

Prior to the gait trials, markers coated with an infrared-light reflecting layer are placed on relevant anatomic landmarks of the subject. A six-camera (VICON) system with a scan rate of 50 Hz was used for determining the 3D marker position. Two subsequent Force Plates with a scan rate 250 Hz were used to simultaneously record the Ground Reaction Forces during walking.

Figure 2-7: A subject with the marker set used for the measurements.

With the recorded marker position- and Ground Reaction Force data as input, the anatomic angular joint kinematics (rotations, velocities, accelerations) and kinetics (torques, powers) were estimated using the procedures described by Koopman.^{10, 11}

Figure 3-6 shows the marker placement. The shoulder markers are placed on both the left and right *Acromio-Clavicular* joints. The hip markers are placed on the both the left and the right *Anterior Superior Iliac Spine*. The sacrum marker is approximately placed between the *sacrum* and the *L1 vertebra*, such that the hip and the sacrum marker(s) are in a plane, which is horizontal to the ground for standing upright. The thigh markers are both placed anterior

laterally on the thigh. The knee markers are placed on both the left and right *Lateral Femoral Epicondyles*. The tibia markers are placed anterior laterally on the lowerleg. The ankle markers are both placed on the lateral *Maleoli*. The toe markers are placed on the 5th *metatarsal heads*. Finally the heel markers are placed posterior on the *calcaneus* (shoe). In case of an unilateral transtibial amputation, the markers on the prosthetic part of the affected side are generally placed such that the placement approximately concurs with the sound limb marker placement.

3 The Highly Efficient Energetic Locomotion Foot

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3 The Highly Efficient Energetic Locomotion Foot

Healthy walking is a dynamic process during which gravitational, kinetic and elastic energy are continuously exchanged in an extremely efficient manner. To maintain this process, energy is transferred from the left -to the right leg (via the trunk) and vice-versa during each WC. If -after a transtibial amputation- the amputee is fitted with a lowerleg prosthesis that does not sufficiently support or rather complement this dynamic walking mechanism, an asymmetry in support and loading of remaining structures is introduced, which increases with increasing walking speed. Literature shows that the prosthetic walker tries to compensate for -or reduce this asymmetry with excessive movements and loading of the remaining structures on both the sound and affected side, resulting in increased energy consumption, early fatigue and decreased walking comfort. Beside the immediate walking discomfort and poor cosmetics, the continuous asymmetric pelvic support increases chance on lower back pain as a secondary disability in the long term.^{6, 22} Furthermore the overall increase in sound side loading increases the risk on early degenerative arthritis.^{3, 11, 22, 30} Although not directly supported by literature, it seems that a similar prospectus can be given for the affected side due to the continuous increased co-contraction of remaining musculature, extra (over) straining the affected knee joint.

Gait studies suggest that the asymmetries in TT-amputee gait walking are the result of current prosthetic feet and their—more or less- rigid ankle joint function. This chapter focuses on the design of a Highly Efficient Energetic Locomotion foot (HEELfoot) that aims to improve current prosthetic walking though re-establishing part of the typical anatomic ankle-foot characteristics occurring during walking at SWS.

In section 3.1 the typical anatomic –and current prosthetic ankle-foot characteristics are discussed more closely. On the basis of these characteristics and TT-amputee gait literature (previous chapter) the design goals of the HEELfoot are explained. Section 3.2 discusses the design synthesis of the HEELfoot. Section 3.3 discusses the possible advantages and the design of an additional coupling between the anatomic knee and the affected ankle joint. Finally, section 3.4 describes the total HEELfoot from concept to prototype.

3.1 Current ankle-foot characteristics and project goals

Since the anatomic foot is the part of the human body that is to be replaced with a prosthetic device after a TT-amputation, the typical anatomic -and current prosthetic ankle-foot characteristics during walking at SWS are discussed more closely in this section. By comparing these two characteristics the main differences can be identified and –with the knowledge of existing TT-amputee gait study reports- suggestions can be made toward

improving current prosthetic feet. Figures 3-1 and 3-2 respectively show typical anatomic and prosthetic ankle characteristics of subjects walking at SWS. The occurring rotation about the ankle joint and the corresponding ankle torque are plotted in one graph instead of two separate graphs as is customary (against the %WC) for explanatory reasons. The vertical axis shows the ankle torque, where torque in plantarflexion direction is positive. The horizontal axis shows the corresponding ankle-angle where a plantarflexion rotation is negative. It should be noted that these graphs merely serve the purpose of identifying the general shape, similarities and differences of prosthetic and anatomic ankle characteristics.

Figure 3-1: Typical anatomic ankle-foot characteristics for normals walking at SWS, considered in the sagittal plane with: a) The different phases of a foot during stance. b) The ankle torque (vertical axis) and the corresponding angle (horizontal axis) which occur during the stance phases. Plantarflexing ankle torques and dorsalflexing ankle rotations are considered positive.^{34, 35}

Figure 3-1 shows that a normal establishes a plantarflexing rotation of the foot about the ankle joint while exerting a net dorsalflexing torque just after HS. This is to establish a quick and

controlled Foot Flat (FF) as the body weight is shifted onto the leg. From FF the dorsalflexing torque gradually changes into a plantarflexing torque while the lowerleg establishes a dorsalflexing rotation relative to the foot. With enough plantarflexing torque build up about the ankle joint the heel is lifted. With HO established, the dorsalflexion initially increases while the plantarflexing torque builds up even more. During the subsequent plantarflexion a maximum in plantarflexing torque is achieved after which this torque declines rapidly with increasing plantarflexion of the foot. At TO, the ankle torque is approximately zero while the foot is in plantarflexion relative to the lowerleg, with the (big) toes approximately flat on the floor. The graph shows that mechanical energy is generated about the ankle joint (because of the counter clockwise torque characteristic).

Figure 3-2: Typical sagittal plane prosthetic foot characteristics of a TT-amputee for walking at SWS (SACHfoot): a) The different phases of the foot during affected stance. b) Shows the ankle torque (vertical axis) and the corresponding angle (horizontal axis) which occur during the stance phases. Plantarflexing ankle torques and dorsalflexing ankle rotations are considered positive.

In comparing the general shape of typical normal ankle-foot characteristics with typical prosthetic ankle-foot characteristics of a TT-amputee walking at SWS (Fig.3-2) it is seen that the two do not differ much from HS until the maximum plantarflexing ankle torque. After that however, it is seen that where the anatomic foot establishes plantarflexion relative to standing upright, the prosthetic foot barely flexes back to neutral. This is typical for current prosthetic feet since all are basically –leaf- springs (some with excellent damping qualities and some fairly rigid) that can only flex back to their neutral position after deformation.

The gait studies discussed in Chapter 2 indicate that incorporating more normal –anatomic- ankle characteristics, occurring from approximately HS until the maximum anatomic ankle torque, to CF is beneficiary for TT-amputee walking: It is seen that allowing for a compliant plantarflexion rotation about the ankle joint during affected heel strike enables an early and controlled, stable FF, improving the prosthetic walkers' sense of stability during early stance. Also allowing for a more normal dorsalflexion rotation and torque about the affected ankle seems to slightly improve the 3C-level of the prosthesis. However, in walking with these types of prosthetic feet an excessive straining of remaining structures remains (sometimes even increases), as the prosthetic walker tries to compensate the lack of affected plantarflexion action during the end of affected stance. This is probably levelling down the 3C-level and increases secondary health risks in the long run.

It seems that adding a more normal plantarflexion action to current prosthetic feet during late affected stance is beneficiary for TT-amputee walking: The plantarflexion of the foot increases the affected leg-length and generates enough energy to enable a smooth progression of the center of gravity as energy is being transferred from affected to the sound side, via the trunk. This makes the excessive straining of remaining structures redundant while enabling a more symmetric support of the trunk. It can even be argued that the lowerleg elongation increases the affected double support time, allowing for more time to transfer energy from the affected to the sound leg, thereby reducing peak residual limb loading.

The HEELfoot aims to increase the 3C-level by enabling a more normal late stance plantarflexion ankle-foot action. This brings forth the following –interrelated- important issues concerning the prosthetic foot design:

- How can the plantarflexing rotation be restored in a prosthetic foot?
- What stiffness characteristic should the prosthetic ankle joint have to enable more normal affected ankle torques to be established with the corresponding plantarflexing rotation (for correct energy storing and releasing)?

Since the aim of the prosthetic design is raising the 3C-level, the three C's: Comfort, Control, and Cosmetics have to be continuously reviewed during the design process.

3.2 A Passive Plantarflexing Ankle Foot Mechanism

Figure 3-3: Two possible switch controlled plantarflexing foot concepts. a) Makes use of the plantarflexing rotation of the foot during weight acceptance. b) Makes use of the relative segment rotations of the foot as the ankle joint drops under loading. The working of both concepts is shortly explained in the main text.

During this project human walking is assumed to approach a ballistic process. This seems reasonable since the human body strives to minimize metabolic energy expenditure. This is also seen in human walking where the design of the lowerlimbs is such that activation -and contraction conditions of musculature are near isometric or eccentric, minimizing the necessity for inefficient concentric muscle contractions.^{7, 31} Muscle functionality can then basically be approached as that of a passive springs with variable stiffness and spring length being activated during certain parts of the walking cycle. Therefore it should be possible -and perhaps even be a necessity to design a passive plantarflexing prosthetic device where the -mechanical- energy stored (absorbed) and released (generated) is taken from and given back to the total walking mechanism during certain parts of the walking cycle, of which the total mechanical energy level (sum of potential and kinetic energy) remains approximately constant.^{16, 31} Beside that, with the current state of technology, an actuator-controlled device that enforces plantarflexion of the foot will become too heavy, as the demanded peak actuator power would approximate 250 Watts (for a person of 80kg.). It was hypothesized that -in conception- a passive plantarflexion device is beneficiary for prosthetic walking, not only from an anthropomorphic

point of view but also for simplicity of construction, weight reduction, and mechanical efficiency of the device. Actuators or controllers can always be added afterwards, should this prove to be necessary (e.g. for walking at different speeds).

Figure 3-3 shows two examples of passive switch-based plantarflexing foot mechanisms. The first mechanism consists of a lowerleg, a foot and a toe segment (Fig.3-3a). In this mechanism the foot and the lowerleg segment are coupled with a flexible element just after the foot has established the maximum plantarflexion rotation relative to the lowerleg during weight acceptance. The flexible element allows for subsequent dorsalflexion rotation and ensures that the same plantarflexing rotation established during HS is established during TO. After TO the foot and lowerleg are uncoupled to ensure sufficient toe clearance with the ground during swing. However, on average a maximum plantarflexion of a mere 5° is established during weight acceptance^{34, 35}, which is substantially lower than the amount of plantarflexion required during TO ($\sim 16^\circ$).

Figure 3-3b shows a prosthetic foot mechanism that consists of a heel, a forefoot, and a toe-segment of which the heel and the toe are connected with a plantar spring. The plantar spring allows for the ankle joint to drop during weight acceptance, causing a plantarflexion rotation of the heel relative to the lowerleg. An assumed maximum ankle-drop of approximately 10mm. during weight acceptance causes the heel to plantarflex approximately 9° relative to the lowerleg. By coupling the heel -and the lowerleg segment with a flexible element at that maximum ankle-drop, the same plantarflexion rotation will be established during TO. However, although the amount of plantarflexion that can be established has increased, it remains substantially lower than the amount of plantarflexion required during TO.

Aiming to re-establish a plantarflexing rotation of the prosthetic foot for ‘through the upperleg’ amputees, Andrae² proposed to temporarily- couple the rotation of knee joint with the rotation of the ankle joint (Fig.3-4). During swing the ankle is uncoupled from the knee using a switch and a relative weak spring lifts the foot. By choosing the correct angular transmission ratio it should be possible to establish enough plantarflexion during TO using this principle.

Probably more switch-based concepts are possible, all using different signals or movements for switching and establishing plantarflexion (e.g. gravity, position or direction GRF under foot, relative segment angles).^{10, 20, 23} With the application of switches however it is difficult to guaranty a fail-save action since the switch itself is a potential hazard for mechanism failure. Certainly some kind of algorithm can be found that minimizes the chance on failure of the switch during walking. However, the addition of switches, controlled by controllers, which can again be controlled, tends to undermine design simplicity and adds to prosthetic weight. Better would be a prosthetic foot that establishes plantarflexion during TO without the aid of switches (Fig.3-5).

Figure 3-4: Prosthetic mechanism that aims to re-establish plantarflexion of the prosthetic foot for 'through the upperleg' amputees. This is achieved by coupling the knee joint rotation with the ankle joint rotation during stance. A switch is used for coupling and uncoupling of the knee and ankle joint. a)/ b) With the knee and ankle rotations coupled a knee flexion rotation causes an ankle plantarflexion rotation. c) With the joint rotations uncoupled the foot is lifted by a weak spring.²

Figure 3-5 shows a prosthetic foot mechanism that consists of a heel, a forefoot and a toe segment, interconnected with hinge joints. A coupler segment couples the heel –and the toe segment. In that manner the rotations of the toe –and the heel segment relative to the forefoot segment are coupled and with the correct linkage configuration a dorsalflexing rotation of the toe relative to the forefoot will cause a plantarflexing rotation of the lowerleg relative to the forefoot (Fig.3-5a). To allow for energy storing and releasing during walking an extra lowerleg segment is added to the linkage mechanism. The lowerleg segment is connected to the heel segment with two compression springs (zero tension force can be exerted). The heel spring is to store the relative small amount of energy needed to establish a quick compliant FF during weight acceptance (plantarflexion rotation while exerting a dorsalflexion torque about the ankle joint). The plantar spring is to store the energy from the following dorsalflexion rotation while exerting a plantarflexing torque about the ankle.

Figure 3-5: a) The rotation of the heel relative to the forefoot and the rotation of the toe relative to the forefoot can be coupled such that a dorsalflexion rotation of the toe relative to the forefoot causes a plantarflexing rotation of the heel relative to the forefoot. b) As a) but with extra lowerleg segment, which is connected to the heel segment with two compression springs (A heel spring and a plantar spring), to enable energy storing and releasing during ambulation.

The –aimed- working of the mechanism is set out in five different phases as shown in Fig.3-6. **(a)**: The neutral segment positions when the foot is unloaded during swing. These relative positions are to be maintained by means of a weak spring. **(a→b)**: From Heel Strike to Foot Flat, the heel spring is compressed and a dorsalflexing torque is build up about the ankle while a plantarflexing rotation is established of the lowerleg segment relative to the heel segment. **(b→c)**: Just after Foot Flat the lowerleg segment commences dorsalflexing relative to the forefoot and the heel segment, causing the heel spring to relax and –eventually- the plantar spring to compress. This causes the dorsalflexing ankle torque to be converted into a plantarflexing torque. With enough plantarflexing torque built up about the ankle joint, Heel

Off will be established. As can be seen two ‘routes’ are possible from b to c: The dotted line concurs more with the average ankle characteristic shown in Fig.3-1 as a plantarflexing torque is exerted when the ankle rotation is zero. Such a characteristic could be achieved by using a – partly- compliant heel segment, which would result in an ankle-joint-drop under loading, causing a backward rotation of the lowerleg segment (much like the SACH feet). As this would mean that the amputee would experience a backward pulling ankle torque in standing upright it is questionable if such a characteristic would be beneficiary for most TT amputees. Preferent is a characteristic where –in conception- no ankle torque is exerted with an ankle rotation equal to zero (the solid line). Should the ‘dotted line’ ankle torque be preferred for standing upright, it can also be achieved by mounting the foot slightly in plantarflexion relative to the shank ($\pm 3^\circ$).

(c→d): After Heel Off the linkage mechanism causes the heel segment to plantarflex relative to the forefoot segment whereas the occurring ankle torque causes the plantar-spring to compress even more, resulting in a dorsalflexing rotation of the lowerleg segment relative to the heel segment. This initially results in a net dorsalflexing rotation of the lowerleg segment relative to the forefoot segment. Approximately at (d) however, the plantarflexion rotation of the heel relative to the forefoot overcomes the compliant dorsalflexion rotation of the lowerleg relative to the heel, resulting in a net plantarflexion rotation relative to the forefoot. **(d→e):** During unloading, both the relaxing of the plantar-spring and the linkage mechanism action cause a plantarflexing rotation of the lowerleg segment relative to the forefoot, resulting in a plantarflexing rotation of the lowerleg relative to the forefoot at TO. **(e→a):** As soon as the foot is unloaded, the links are pulled back to their neutral position by a weak spring.

In conception this working seems sound: A linkage mechanism enables lowerleg elongation by means of plantarflexion of the forefoot relative to the lowerleg and a flexible joint -or link- enables energy storing and releasing during walking. By varying the linkage configuration the amount of plantarflexion can be varied and the timing of energy storing and releasing can be optimised by –simultaneously- varying the flexible joint stiffness. Since the amputee controls the torque exerted about the affected ankle, the amputee himself controls the timing of energy storing and releasing. Finally it is noteworthy that, although only considered as a 1dof – mainly- sagittal plane joint in this thesis, a flexible joint that allows for 2dof or 3dof can also be mounted to the described linkage mechanism (See the section ‘statically gravity balancing for rotations about a fixed pivot’).

Figure 3-6: The hypothesized working of the prosthetic foot concept shown in Fig.3-5b. Figures a) to e) visualize the different phases or positions of the foot during a walking cycle. Figure f) gives the outline of the aimed prosthetic ankle joint characteristic; plantarflexing ankle torques and dorsalflexing ankle rotations are considered positive (principle patented, nr.:NL1018616, PCT-patent pending).

3.2.1 Mechanism Ankle Joint Stiffness

Basically, with a four bar linkage to establish the plantarflexion with, the remaining question is what initial ankle joint stiffness of the foot could best be chosen for the mechanism to function properly during walking. Thus, since the ankle joint consists of two compression springs, both subsequently active during walking, two different initial ankle joint stiffness should be determined: One for plantarflexion and one for dorsalflexion of the lowerleg segment relative to the heel segment causing compression of the heel spring and the plantar spring respectively. Since the actual aim of the new prosthetic foot is to improve on current TT-amputee walking by means of restoring a more natural ankle joint function, anatomic ankle joint function was chosen as a guideline for determining initial ankle joint stiffness.

Figure 3-7: Determination –initial- ankle joint stiffness for plantarflexing rotations of the lowerleg segment relative to the heel segment.

In doing so, the ankle joint stiffness for plantarflexion of the heel segment relative to the lowerleg segment was chosen such that the maximum measured dorsalflexing torque just after Heel Strike occurs at the maximum measured plantarflexion rotation just after heel strike as measured by Winter^{34, 35} for normals walking at SWS (a→b, Fig.3-6f). With the heel spring a linear compression spring, the spring stiffness can be determined according to (Fig.3-7):

$$C_h = \frac{M_b}{r_m \sin \beta \cdot (l_a - l_b)} \quad 1)$$

With C_h the heel spring stiffness.

To establish a smooth ambulation the plantar spring characteristic is of great importance. The spring should not be too weak to allow for sufficient support in standing or walking. If the spring is too stiff however, the amputee will not be able to establish a smooth ambulation and peak stump loads and asymmetry are increased during walking. With as many different ankle joint characteristics exist as there are amputees it is impossible to find one all defining measure for determining ankle joint stiffness. However, to have some sort of measure to set initial ankle

joint stiffness for different people with different measures, it was decided to use gravity balancing as a guideline. Therefore the initial ankle joint stiffness is set such that gravity acting on the Centre of Gravity (COG) of a subject standing on the affected leg and rotating about the prosthetic ankle joint -in an upright position- is (approximately) balanced by a counteracting torque M_p exerted by the ankle joint (Fig.3-8).^{25, 26}

Figure 3-8: Ankle joint with a rotational stiffness such that the gravity acting on the COG -at a distance R from the ankle joint- is statically balanced.

3.2.2 The Mechanism Behaviour from HO until TO

From Heel Strike to Heel Off, the working and built up of the proposed concept foot are similar to prosthetic feet currently available on the market. There is not much worry about the working of the proposed concept during this period (with the necessary adjustment possibilities incorporated in a future prototype foot). From Heel Off to Toe Off however, the linkage mechanism activates to re-establish the lowerleg elongation through plantarflexion of the forefoot segment. Since the prosthetic foot working during this phase of the Walking Cycle was entirely new, some kind of ‘pre-prototype’ validation was done to ensure that the hypothesized working was at all possible. This was done numerically with typical anatomic ankle-foot characteristics occurring during walking at SWS as the reference frame. By comparing the mechanisms behaviour established under ‘anatomic loading’ to the anatomic ankle-foot characteristics of a normal subject walking at SWS, it was assessed whether or not the proposed mechanism could approach anatomic ankle characteristics in the period from

Heel Off to Toe Off. The dataset used for this validation was that of a subject with typical normal gait characteristics (see also Chapter 2) and was probably unique in the fact that his ankle characteristic actually approached the one derived from the -averaged- Winter data set (see also Fig.3-1 and section 3.6.1 of the Appendix).

Figure 3-9: Net loads acting on the ankle joint (considered in the sagittal plane) from Heel Off to Toe Off. F_{gri} and M_{ai} are the force and torque acting on the ankle joint respectively at certain points in time i between Heel Off and Toe Off. The angles α_i and γ_i are the rotations of the lowerleg segment and the forefoot segment relative to standing upright respectively. a) The position of the forefoot and lowerleg at HO. b) The position of the foot and lowerleg between HO and TO. c) Geometrical measures of the foot for standing upright.

Figure 3-9 visualizes the net loads acting on the ankle joint and the relative segment rotations occurring during normal walking at Heel Off (Fig.3-9a) and at a certain % i of the Walking Cycle between Heel Off and Toe Off (considered in the sagittal plane, Fig.3-9b). The force F_{gri} acting on the ankle joint at an angle ϵ_i to the vertical has approximately the same value as the

measured Ground Reaction Force (inertia and gravity forces of foot considered negligible during stance) while a net torque M_{ai} acting about the ankle joint is then demanded for equilibrium. Furthermore it is seen that the lowerleg segment and the forefoot segment are rotated through an angle of α_i and γ_i respectively, relative to standing upright. The difference of α_i and γ_i is the rotation ψ_i of the forefoot segment relative to the lowerleg segment (same as the anatomic ankle rotation discussed in the previous sections):

$$\psi_i = \alpha_i - \gamma_i \quad 2)$$

With $\psi_i > 0$, the forefoot has established a dorsalflexing rotation relative to the lowerleg.

With $\psi_i < 0$, the forefoot has established a plantarflexing rotation relative to the lowerleg.

For the mechanism discussed, the length of the forefoot (r_3), the toe (r_t) and the heel (r_h) segments of the linkage mechanism are estimated directly from anatomic foot landmarks of an upright standing person (Fig.3-9c): The forefoot length and the start angle γ_s of the forefoot segment relative to the horizontal are estimated by measuring the distance between and the vertical displacement Y of the centres of the medial maleolus and the medial metatarso-phalangeal joint respectively. The length of the toe segment is estimated by measuring the distance between the centre of the medial metatarso-phalangeal joint to the end of the big toe (person standing upright). Finally the length of the heel segment is estimated by measuring the distance between the centre of the medial maleolus to the most posterior part of the heel still in contact with the ground. The length and initial angular positions of the remaining links in the linkage mechanism are variable.

With the main segment lengths and initial positions of the proposed linkage mechanism estimated according to anatomic landmarks of the subjects foot, it seems reasonable to assume that the proposed mechanism can support this gait pattern provided the mechanism establishes –approximately- the same relative segment rotations with the same –anatomic- loads acting on it (assuming that approximately the same net joint torques and rotations are established at the hip and knee during walking). To validate this, first the anatomic ankle force F_{gri} is imposed on the mechanisms ankle joint at an angle ε_i to the vertical while the corresponding anatomic forefoot rotation γ_i is enforced on its forefoot segment (see Fig.3-10). For the linkage mechanism to remain in equilibrium under these conditions, a torque M_{2,a_i} acting on link r_2 (=part of heel segment) and about the ankle joint is demanded. In this case this means that the line of action of F_{gri} must pass through the Instantaneous Centre of Rotation of link r_2 and the toe segment (see section 3.6.2. of the Appendix).

The torque M_{2,a_i} causes a clockwise rotation $\alpha_{Mi} = \alpha_s - \alpha_{\gamma_i}$ of the lowerleg segment relative to the heel segment (while compressing the plantar spring) where the enforced rotation γ_i of the forefoot segment causes a counter clockwise rotation $\Delta\beta_i = \beta_s - \beta_{\gamma_i}$ of the heel segment relative to the forefoot segment. With the plantar spring stiffness chosen such that gravity

acting on the Centre of Gravity is balanced (Fig.3-8), and the –remaining- linkage configuration chosen, the rotation ψ_{mi} of the mechanisms' lowerleg segment relative to the unloaded start positions can be determined according to:

$$\psi_{mi} = \alpha_{Mi} - \Delta\beta_i \quad 3)$$

With $\psi_{mi} > 0$, dorsalflexion of the forefoot segment relative to the lowerleg segment.

With $\psi_{mi} < 0$, plantarflexion of the forefoot segment relative to the lowerleg segment.

For this –passive- prosthetic foot mechanism to be able to support this walking pattern, both the mechanisms' ankle torque $M_{2,ai}$ and the mechanisms' relative forefoot rotation ψ_{mi} should approximate the anatomic ankle joint torque M_{ai} and its corresponding relative forefoot rotation ψ_i .

Figure 3-10: Proposed mechanism with the measured ankle force imposed on the ankle joint while the simultaneously occurring rotation γ_i enforced on the forefoot segment. An ankle torque $M_{2,ai}$ of the mechanism is then demanded for equilibrium. a) Foot in unloaded –start- position. b) Link r_3 (forefoot segment) rotated through an angle γ_i with the corresponding external force F_{gri} enforced on the ankle joint at an angle ϵ_i to the vertical and the torque $M_{2,ai}$ that is demanded for equilibrium. The heel spring is not displayed because it is inactive throughout the considered period (HO to TO).

The relevant measures of the subject are shown in Table 3-1 together with the linkage mechanism dimensions that have been used for this consideration. It is seen that two mechanisms (A and B) have been considered. Both are identical except mechanism A has a

link length $r_4 \neq 0$, whereas mechanism B has link length $r_4 = 0$. With $r_4 = 0$, no plantarflexion action is possible as the heel, the forefoot and the coupling segment form one rigid element, only allowing for a rotation about the toe joint (As is the case for current prosthetic ankle foot mechanisms).

Subject	77kg	$R=1.06m$	(Also see Fig.3-8 and 3-9)				
Mechanism A	$r_3=0.14m$	$r_i=0.075m$	$r_2=0.055m$	$r_4=0.02m$	$\gamma_s=28^\circ$	$\rho=0^\circ$	$\xi=0^\circ$
Mechanism B	same	same	same	$r_4=0$	Same	same	Same

Table 3-1: Relevant measures of the subject and the proposed mechanisms that were used for the numerical validation.

Figures 3-11a and b, show the calculated ankle torque $M_{2,ai}$ with the corresponding rotation ψ_i of the forefoot segment relative to the lowerleg segment (relative to the unloaded start position of the linkage mechanism). The reader is referred to the Appendix (section 3.6.2) for a more elaborate discussion on the kinematics and kinetics of the mechanism.

The torques M_{2,ai_A} and M_{2,ai_B} are the ankle torques needed for equilibrium of mechanism A and B respectively. It is seen that the ankle torque M_{2,ai_A} of mechanism A approximates the anatomic torque M_{ai} about the ankle quite reasonably (Fig.3-11a). The torque needed for equilibrium of mechanism B (M_{2,ai_B}) is lower than for mechanism A. This is because the Instantaneous Centre of Rotation of the lowerleg segment relative to the toe segment moves forward and downward during ambulation relative to mechanism B in which the heel, the forefoot and the coupler segment rotate as one element about a fixed pivot, ‘the toe joint’ (Fig.3-12).

Figure 3-11b shows the anatomic and the mechanisms’ forefoot segment rotations relative to the lowerleg. It is seen that the rotation of the forefoot relative to the unloaded lowerleg segment position is again approached quite well with mechanism A. Since mechanism B cannot establish plantarflexion, ψ_{mi} progressively deviates from the anatomic ankle angles.

It can therefore be concluded that the ankle/ forefoot characteristics of the proposed linkage mechanism in combination with an approximate gravity balancing ankle joint (mechanism A)

can approximate the typical anatomic ankle foot characteristics of a subject walking at SWS in the period from Heel Off to Toe Off.

Figure 3-11: Anatomic and mechanism ankle characteristics at the different %WC from HO to TO with: a) The anatomic (M_{ai}) and mechanism (M_{2,ai_A} and M_{2,ai_B}) ankle torques on the vertical axis with the corresponding %WC on the horizontal axis. b) The anatomic (ψ_i) and mechanism (ψ_{mi_A} and ψ_{mi_B}) ankle angles on the vertical axis with the corresponding %WC on the horizontal axis: Plantarflexing ankle torques and dorsalflexing ankle rotations are considered positive.

Figure 3-12: The path of the ICR ($r_4 \neq 0$), as a function of the rotation γ_i of the forefoot. For $r_4=0$, the foot rotates about a fixed pivot.

3.2.3 Conclusions

It seems that adding a more normal plantarflexion action to current prosthetic feet during late affected stance seems beneficiary for TT-amputee walking: The plantarflexion of the foot increases affected leg-length and generates enough energy to enable a smooth progression of the Center Of Gravity as energy is being transferred from affected to the sound side, via the trunk. This makes the excessive straining of remaining structures redundant while enabling a more symmetric support of the trunk. It can even be argued that the lowerleg elongation increases the affected double support time, allowing for more time to transfer energy from the affected to the sound leg, thereby reducing peak residual limb loading. It is hypothesized that such a mechanism will substantially improve the 3C-level during walking.

A passive mechanism without any actuators or switches has been proposed for establishing this plantarflexion action. Passive mechanisms tend to increase simplicity of design, ease of construction and the fail safety of the mechanism, and reduce prosthetic weight. The energy needed for establishing plantarflexion is then taken from and given back to the total prosthetic anatomic system in which human walking is considered to be a ballistic process.

The proposed mechanism consists of a four bar linkage in combination with a flexible link or joint. The flexible link is chosen such that the prosthetic ankle joint has a rotational stiffness such that the gravity acting on the COG of a subject standing in an upright position on the affected leg is statically balanced for dorsalflexing rotations about the prosthetic ankle joint (Fig.3-8).

The working of the mechanism has been compared –numerically- with typical sagittal plane anatomic ankle-foot characteristics occurring during normal walking at SWS as a ‘pre-prototype’ validation. From that comparison can be concluded that the mechanisms behaviour under anatomic loading conditions can approach typical anatomic ankle-foot characteristics. Since it has been hypothesized that incorporating typical anatomic ankle-foot characteristics occurring during walking into a prosthetic foot can be beneficiary for a prosthetic walker, this result justified the further development of the proposed concept into a prototype foot, which can be tested during clinical trials.

It is recognised however, that this way of numerically ‘pre-assessing’ the performance of a new type of prosthetic foot prior to clinical trials is limited since the human factor (residual limb pain, human control) is not -and with current technology cannot be- accounted for. Furthermore, the prosthetic-anatomic walking mechanism remains different from the healthy walking mechanism. After all, the prosthetic-anatomic walking mechanism has an extra flexible interface (socket/ stump), and lacks certain anatomic structures normally active during walking. Thus, true assessment of the assumptions made in the described design process can therefore only be established on the basis of patient feedback and gait-measurements.

3.3 The Knee Ankle Coupling

The plantarflexing functionality which is established by coupling the rotation of the toe with the rotation of the ankle joint, is -to a lesser extent- also present in the anatomic foot where structures underneath the anatomic foot -that couple the toes with the heel and lowerleg- cause the -initially quite movable- bony structures in the forefoot to tense like a bow when contracted or when the toes are lifted, as the ball of the foot is pulled downward and toward the heel (Fig.3-13). The forefoot combined with these structures then behaves as a stiff leaf-spring like whole, which is necessary to support walking. With the forefoot allowed to rotate freely about the ankle joint, it is seen that plantarflexion of the forefoot relative to the lowerleg is established. Research indicates that up to 10° of the plantarflexion established during TO is caused by this effect, called the windlass effect.^{14, 15, 18, 28, 29}

Figure 3-13: Schematic representation of the effect of the plantar structures and Flexor Hallucis Longus on the bony structures of the foot when contracted or when toes are lifted when the foot is: a) standing on the ground. b) Forefoot allowed to rotate about the ankle joint.

These couplings are called bi-articular couplings because they span two joints. Meaning that the structures like the PS and the FHL are actually poly-articular couplings since more than two joints are spanned if all joints in the anatomic foot are accounted for. This section discusses possible advantages of re-establishing bi-articular functionality of other structures in the lower extremity by means of an orthotic extension of the lowerleg prosthesis, after an amputation has occurred.

3.3.1 Bi-articular couplings in lower extremity

Beside the Plantar Structures in the anatomic foot, several other main bi-articular muscle groups can be distinguished when the whole lower extremity is considered. The gastrocnemius (part of the calf muscle) connects the foot directly with the upperleg enabling simultaneous

knee flexor and an ankle plantarflexor action. The quadriceps directly connect the pelvis with the lowerleg enabling simultaneous hip flexor and knee extensor action. Finally the hamstrings connect the pelvis with the lowerleg enabling simultaneous hip extensor and a knee flexor action. Figure 3-14 shows the joint angles and torques occurring during walking in combination with a picture of the corresponding musculature activity during walking.^{16, 34, 35}

Figure 3-14: The relative joint angles and joint torques with the corresponding activity of the different musculature during a walking cycle.^{16, 34, 35}

Just before HS, the hamstrings and quadriceps are active, creating a net flexion torque about the knee and an extension torque about the hip. Both are needed to decelerate the lowerleg rotation relative to the trunk (80%→100% WC). During HS, Tibialis Anterior activity

increases and a corresponding net dorsiflexing torque about the ankle joint is measured (to ensure a controlled FF; 0→10% WC). Subsequently, quadriceps activity maintains while hamstrings activity decreases, resulting in a net extension torque about the knee joint, preventing the knee from collapsing during weight acceptance (5-27%WC). After FF the Tibialis Anterior activity decreases while calf musculature activity increases and a plantarflexing torque is built up to establish Heel Off somewhere between 20%-40% of the Walking Cycle (10%→60% WC). Somewhere in the period 27%→45% WC the calf muscle activity peaks whereas quadriceps and hamstrings activity is low resulting in a net flexion torque about the knee and a plantarflexion torque about the ankle joint. Subsequently the foot starts plantarflexing while the calf muscle gradually deactivates, maintaining a plantarflexing torque about the ankle joint. At the same time quadriceps activity increases and the knee flexion torque is gradually converted into a knee extension torque (45%→80% WC) to accelerate the lower extremity into swing after Toe Off. During swing Tibialis Anterior activity increases, and a net dorsal flexing torque is measured, needed to lift the foot to establish enough toe clearance with the ground. The other structures show little activity. Just before HS the cycle starts again.

Since a relation between muscle force and measured EMG activity has not yet been identified, it is not possible to determine the contribution of different muscle groups to the net joint torque acting about that joint, making it impossible to determine the contribution of mono and bi-articular musculature during walking. Besides that, a seemingly bi-articular structure can act as two separate mono-articular structures, due to the presence of connective tissue between different structures (Fig.3-15).

Figure 3-15: Muscle that spans two joints. (a) Without connective tissue (b) With connective tissue. Without connective tissue, equilibrium demands equal the tensile forces. With connective tissue, the tensile force in the tendons can differ.

In studying explosive leg extension however, Jacobs¹⁷ concludes that bi-articular structures play an effective role in the power transfer from proximal to distal joints where transferring power from the large mono-articular muscles spanning proximal joints, allows high power

output at the smaller distal muscles. Although in this case explosive leg extensions were studied, other researchers hypothesized similar bi-articular functionality during human walking: Bi-articular structures are then assumed to decrease mechanical and metabolic energy expenditure by transferring energy from seemingly energy absorbing joints to an energy generating joints with less –similar sized- muscles.^{16, 32, 34, 35} Furthermore, it is hypothesized that efficient use of bi-articular muscles enables the lowerleg muscles to fire with minimal change in length, approaching isometric or eccentric conditions for muscle activity rather than inefficient concentric contractions.^{7, 31}

Although none of these studies is truly conclusive regarding the use of bi-articular structures during walking, bi-articular muscle functionality seems beneficiary for walking, especially from a minimal energy expenditure perspective. After all, it does not seem logical that the human body would use two actuators to control two joints when one is enough and metabolic energy consumption could be decreased.

3.3.2 The knee ankle coupling

Assuming bi-articular muscle functionality during walking, it seems a necessity to restore bi-articular structures after amputation if a more symmetrical support of the trunk and more natural distribution of loads (joint torques) on the lower extremities of a TT-amputee are aspired. The necessity of re-establishing bi-articular couplings then depends on the remaining functionality of the (remaining) bi-articular structures after amputation. For instance: A restored coupling between the anatomic hip and prosthetic knee allows for a trans-femoral amputee to have increased control over the prosthetic leg during swing, increase knee stability during the end of the stance phase, and most likely increased sensory feedback from the prosthesis.^{23, 26} For TT-amputees with sufficient hamstrings and quadriceps functionality, only restoring the coupling between the anatomic knee and the prosthetic foot should re-enable more –natural- transferring of energy from the knee to the foot during walking, allowing for the remaining structures to maintain a more natural level of loading after amputation. The hypothesis is that this will increase walking comfort by making the process less strenuous. Furthermore the amputee is expected to have more control over -and sensory feedback from the prosthetic foot during ambulation. The fact that some sort of extra orthotic extension of the current lowerleg prosthesis is needed to re-establish the coupling can be seen as a disadvantage regarding cosmetics and easy donning and doffing of the prosthesis. However, an orthotic extension in combination with a bi-articular coupling also offers several strong extra advantages such as decreased residual limb loading and increased knee stability during affected stance. This increases walking comfort and decreases the chance on health problems in the long term. This is why it was decided to also incorporate an optional compliant coupling between the knee and the ankle joint in the HEELfoot design.

Considering Fig.3-14, bi-articular gastrocnemius action is the most logical from 26%→52% where an ankle plantarflexing torque concurs with a knee flexing torque and the calf musculature display high activity whereas the hamstrings and quadriceps display a decreased activity. During this period a knee flexion torque starts building at 26% of the WC with corresponding knee and ankle rotations of 14° and 6.4° respectively. The maximum knee flexion torque ($\approx 0.27\text{Nm/kg.}$) occurs at 42% of a WC with corresponding knee and ankle rotations of 8° and 9.5° respectively. At approximately 52% WC the knee torque has reduced to zero with corresponding knee and ankle rotations of 17° and 4.7° of respectively. During this period, part of the knee -and ankle joint torque can be controlled with just one muscle group; the gastrocnemius.

Figure 3-16: a) Knee-ankle coupling during standing upright. The mechanism consists of a knee lever r_k , an ankle lever r_a and a shank Sh . The linear g-spring that connects the ends of both levers is just strained with a length L_s in this position. b) Mechanism with the joint rotations occurring at 26%WC enforced on it. The lever length ratio r_k/r_a is chosen such that $L_s=L_{26\%WC}$. c) Mechanism with the joint rotations and the corresponding GRF $F_{gr_42\%}$ occurring at 42%WC enforced on it. d) The GRF $F_{gr_42\%}$ causes an external knee extension torque $M_{k_42\%WC}$ that has to be counteracted by the g-spring with a force F_t .

This hypothesized gastrocnemius action has been approximated with a passive linkage mechanism (Fig.3-16) consisting of a shank, an ankle lever r_a and a knee lever r_k , interconnected with hinges. A linear –gastrocnemius- tension spring (g-spring) connects the ends of both levers, spanning both hinges. In standing upright, the g-spring has a length L_s and is slightly strained (negligible, Fig.3-16a). The link length ratio of r_a and r_k is then chosen such that the g-spring also has that length L_s at approximately 26% of the WC, with relative knee and ankle rotations of 14.1° and 6.4° respectively enforced on the linkage (Fig.3-16b). From that point during the WC, the knee flexion torque builds as the g-spring is stretched. The (linear) g-spring stiffness C_g is now determined such that the maximum anatomic knee flexion torque of 0.27Nm/kg is established at 42%WC with the corresponding relative anatomic knee and ankle rotations of 7.9° and 9.5° respectively, enforced on the links r_k and r_a (Fig.3-16c and d). The reader is referred to the Appendix (section 3.6.3) for a more elaborate discussion on the basic mathematics.

Figure 3-17: The knee and ankle torques (calc_knee and calc_ankle respectively) of the mechanism described in Fig.3-16 from 26%WC to 52%WC when the corresponding anatomic knee and ankle rotations are enforced on the joints. The corresponding anatomic knee and ankle torques (knee and ankle respectively) are also shown.

Having determined the g-spring stiffness and chosen the link lengths, the knee and ankle torques can be determined as function of the relative knee and ankle rotations occurring during a walking cycle. Figure 3-17 compares the anatomic knee and ankle torques with the knee and ankle torques of the mechanism with the corresponding anatomic knee and ankle rotations enforced on the mechanisms' links. This is done from 26%WC to 52%WC since the g-spring is slack during the rest of the WC. It was striking to see that the anatomic knee torque could be

approached so well with a passive linkage mechanism. This -combined with the facts that the g-spring stretched for maximum 5mm and that a human anatomic lever arm ratio $r_k/r_a \approx 1/2$ (Cannatella, Kawahami et. al) was used in this mechanism configuration- at least shows that it is possible for the gastrocnemius to support the knee with isometric muscle contractions, using the tendons as the spring elements. Also it is seen that approximately 36% of the total anatomic ankle torque that is required for equilibrium during normal walking, is supported by the knee-ankle coupling (at 42%WC). Because the knee torque simultaneously causes an ankle plantarflexion torque, re-establishing the knee-ankle coupling enables a -TT- amputee to have more control over the occurring ankle torque from the knee (by extra or less extending the knee) from 26%WC to 52%WC.

3.3.3 Effects knee ankle coupling on residual limb loading

As already stated in this section, the fact that an orthotic extension of the current lowerleg prosthesis is needed to re-establish the coupling can be seen as a disadvantage regarding cosmetics and donning and doffing of the prosthesis. This section shortly discusses the possible advantages of an orthotic extension regarding the residual limb loading.

Figure 3-18 is a side view of a TT-amputee in a forward leaning position. For simplicity, the effect of all forces acting on different areas of the residual limb and the socket are combined to single force vectors. The forward lean demands a net knee flexion torque from the musculature acting about the knee joint for equilibrium (similar to the loading shown in Fig.3-16). With a PTB-socket (Fig.3-18a), this knee flexion torque is counteracted with a couple $F_{cs} * a$. In this case both forces have to be directly supported on the residual limb with one of them acting on the posterior-distal end. However, with an orthotic extension in combination with the knee-ankle coupling, a large part of this counteracting couple can be directly supported on the upperleg, reducing the distal stump loading.

A similar effect of the orthotic extension is seen when the residual limb loading of a TT-amputee is considered in the frontal plane in a position corresponding with the mid-stance phase of a WC (Fig.3-19, modified from Inman¹⁶). It is seen that part of the uncomfortable laterodistal contact force, needed for equilibrium can be supported directly on the medial side of the upperleg. The orthosis then acts in cooperation with the laterodistal socket contact force, providing mediolateral stabilisation of the entire fitting.¹⁶ Such a fitting will most likely increase the sense of mediolateral stability during weight acceptance. Therefore it is well possible that the above normal quadriceps and hamstrings activity to stabilize the knee joint during weight acceptance will reduce, reducing the chance on early degenerative arthritis while further increasing walking comfort.

Figure 3-18: Sagittal view of forces acting on residual limb with a forward leaning TT-amputee. a) Forces on residual limb without knee-ankle coupling. b) Forces on residual limb with the knee-ankle coupling.

Figure 3-19: Frontal view of forces acting on residual limb with the subject in a position corresponding with the midstance phase of the WC. a) Forces on residual limb without orthotic extension. b) Forces on residual limb with orthotic extension.

3.3.4 Conclusions

Although –with current measuring techniques- it is not yet possible to determine the function of bi-articular musculature during walking, several studies hypothesize that bi-articular structures serve to transfer power between joints. In that manner power can be transferred effectively from proximal to distal, furthermore energy expenditure can be reduced because one muscle is used to control two joints and effective use of two joint muscles allows for near isometric contractions during walking.

Assuming bi-articular muscle functionality during normal walking, it seems a necessity to restore this after amputation to re-establish a more symmetrical support of the trunk and more natural distribution of loads (joint torques) on the lower extremities. Considering the hypothesized working of these anatomic structures, this will effectively make walking less strenuous for an amputee. For a TT-amputee with good hamstring and quadriceps functionality, only restoring the coupling between the knee and the ankle will suffice.

The anatomic knee-ankle (gastrocnemius) action is the most logical from 26%WC→52%WC since during the period an ankle plantarflexing torque concurs with a knee flexing torque and

the calf musculature display high activity whereas the hamstrings and quadriceps display a decreased activity.

This anatomic knee-ankle coupling action has been approximated with a passive linkage mechanism (Fig.3-16) consisting of a shank, an ankle lever r_a and a knee lever r_k , interconnected with hinges. A linear –gastrocnemius- tension spring (g-spring) connects the ends of both levers, spanning both hinges. With the mechanism configuration chosen such that it approached anatomic measures, it has been shown that the anatomic knee torque can be accurately approached. Also it is seen that approximately 36% of the total anatomic ankle torque that is required for equilibrium during normal walking, is supported by the knee-ankle coupling (at 42%WC). This shows that it is possible for the gastrocnemius to support the knee with isometric muscle contractions, using the tendons as the spring elements.

The fact that an orthotic extension of the current lowerleg prosthesis is needed to re-establish the coupling can be seen as a disadvantage regarding cosmetics and donning and doffing of the prosthesis. However, -besides the hypothesized reduction of energy expenditure- the knee ankle coupling in combination with the orthotic extension of the prosthesis offers several strong advantages such as; reduction of uncomfortable distal loading on the residual limb, increased frontal plane knee stability, more control of and sensory feedback from the prosthetic foot during walking. Increasing walking comfort and decreasing chance on health problems in the long term. Whether the advantages outweigh the disadvantages can only be assessed by the amputee. It should also be noted that weight bearing forces remain transferred through the ‘residual limb-socket interface’ as is the case with current PTB sockets since it is intended only to relief part of the posterior-anterior –and medial-lateral loading on the residual limb and increase the knee stability on the affected side by transferring them onto the upperleg. This allows for lightweight and cosmetically sound construction of the orthotic extension, especially when compared to current conventional –weight bearing- thigh cuffs.

This is why it was decided to incorporate an optional knee ankle coupling in the HEELfoot design. The averaged knee-ankle characteristics as published by Winter are used to determine the initial knee-ankle coupling configuration (as described in section 3.3.2).

3.4 The Prototype Prosthesis

In the course of this section, the development of the passive plantarflexing prosthetic foot concept (section 3.2) and the knee-ankle coupling concept (section 3.3) into a total prosthetic prototype is shortly discussed. Important issues in the development of prototype were easy adjustability, ease of construction and prosthetic weight (approximately 500gr.).

First two configurations of the presented plantarflexing foot concept will be discussed. Subsequently the load set used to dimension the prototype prosthetic foot is discussed. Finally, the total prosthetic device, inclusive knee-ankle coupling will be presented.

3.4.1 The ankle –and foot mechanism

Regarding the prototype prosthetic foot, the main question was how a -mechanically simple-gravity balancing ankle device could be constructed. Having found such a mechanism the foot can be dimensioned.

Figure 3-20: A Mass rotating about a fixed pivot that is approximately balanced with one linear technical spring. a) Mechanism in neutral position with spring relaxed. b) Mass rotated through an angle θ . c) As b) with spring force F_s and gravitational force mg acting on the mass displayed.

The section on gravity balancing discusses several mechanisms that enable static balancing of the gravity acting on a mass rotating about a fixed pivot by means of springs. One of these mechanisms is the one shown in Fig.3-20. This mechanically simple 1dof mechanism approximately statically balances gravity acting on the mass rotating at a distance R about a

fixed pivot with a single linear tension spring. It is explained that with the mechanism dimensions chosen according to the dimensions of the prosthetic foot concept shown in Fig.3-11 ($a=50\text{mm}$, $b=140\text{mm}$, $\phi_0=62^\circ$), the maximum deviation from perfect static balance is a mere 1.5% within the operating range $\theta=0\rightarrow10^\circ$. Bearing in mind that gravity balancing is used as a guideline for determining the initial prosthetic ankle joint stiffness, it is fairly save to say that such a deviation poses no threat to the mechanisms working during clinical trials. The spring stiffness of the mechanism is then determined according to:

$$C_s = \frac{F_{s_req_10^\circ}}{l_1 - l_0}$$

With $F_{s_req_10^\circ}$ the spring-force required for equilibrium with the mass rotated through an angle $\theta=10^\circ$.

Although other –quasi 3dof- balancing mechanisms are also discussed in section 2, it was decided to incorporate this –approximate- balancer in the prosthetic foot. This was mainly because of the ease of construction of the mechanism, but also to reduce the number of unknowns during first clinical trials. The reader is referred to the gravity balancing section for a more elaborate discussion on different gravity balancing mechanisms.

To incorporate the approximate balancing ankle joint in the total foot concept, two mechanism configurations were considered: Configuration A (Fig.3-21a) that is basically the same as the foot mechanism already discussed. Although mechanically reasonably simple, mounting the spring underneath the foot as shown in configuration B (Fig.3-21b) allows for more compact building, effectively reducing foot height (leaving more room for mounting the foot underneath the socket) while more space is available to build in the spring element, without complicating construction. In configuration B this has been achieved by replacing the rigid coupler segment with a linear flexible element. Furthermore, that flexible element directly couples the toe segment with the lowerleg segment instead of the heel segment, as was the case for configuration A.

However, with the coupler segment in configuration B now flexible, the path of the Instantaneous Center of Rotation changes slightly under loading when compared to configuration A (Fig.3-21c and d). This in turn causes a change in the ankle torque required for equilibrium after Heel Off. To ensure a similar working as configuration A, the mechanisms ankle-foot characteristics have been compared with anatomic ankle-foot characteristics -under normal walking conditions loading, assuming a Center of Gravity balancing ankle joint- as described in section 3.2.2. This analysis reveals a maximum deviation of 0.4% from the results achieved for mechanism A (see Appendix: section 3.6.2). Taking into account the constructional advantages of configuration B over configuration A, it was decided to further develop configuration B into a prototype.

Figure 3-21: Two different configurations of the prototype prosthetic foot with built in – approximate balancer shown in Fig.3-20. a/ c) Configuration A which is basically a linkage mechanism with flexible joint discussed previously. b/ d) Configuration B in which the flexible joint is replaced with a flexible coupler segment. With the coupler segment flexible, the path of the ICR changes relative to a rigid coupler segment in combination with a flexible ankle joint. It is seen that rotating the lowerleg relative to the forefoot causes the Instantaneous Centre of Rotation to move along the extension of the forefoot segment in case of a flexible coupler segment (d) where it remains in place with a rigid coupler segment (c).

3.4.2 Dimensioning of Prototype Prosthesis

With the basic set-up of the prototype now known, the next step on towards a prototype is assessing the prototype loading during use and subsequently dimensioning the prototype such that it can withstand this loading. For this the static structural testing procedure for prosthetic lowerlegs as set out in the ISO-103028 protocol was used. It is noteworthy that the ISO protocol describes both dynamic and static testing procedures. However, it was decided to initially test the prototype according to the static procedure because the TT-amputee initially only walks one or two half hours a week under laboratory conditions). Using ISO protocol, the loads on the different links were determined and dimensioned such that maximal yield stresses were not exceeded. Beside the maximal allowed link stresses, total weight of the foot (approximately 500 gr.) and ease in construction were boundary conditions in shaping the

prototype foot. Considering that the prototype dimensioning was done using basic strength-stress-shear formulas, only the ISO-loading conditions and the resulting mechanism are discussed.

Figure 3-22: Static testing prototype according to ISO 103028: a) Prototype in unloaded position. b) As a) with whole foot rotated 20° backward, subsequently the lowerleg link is fixed to the ground and a load F_l is exerted on the heel. c) As a) with forefoot rotated through 20° forward relative to neutral position. Subsequently the lowerleg link is fixed to the ground and a load F_l is exerted on the forefoot.

According to the ISO protocol two static tests have to be conducted (Fig.3-22): First the foot as a whole is rotated backward through an angle of 20° relative to neutral. Then the lowerleg segment is fixed to the ground and a load $F_l=2240N$ is applied to the heel according to Fig.3-22b. Secondly the forefoot is rotated forward through an angle of 20° relative to the toe and the same load $F_l=2240N$ is applied to the toe (Fig.3-22c). The mechanisms springs have been replaced with rigid blocks during testing; this because the loads occurring during walking at self-selected walking speed are much lower than the static ISO-loading that was prescribed.

Strictly following the ISO protocol, the line of action of this load lies in the sagittal plane, causing only (favourable) pressure and tension loads in the linkage mechanism. During normal walking however tangential forces in the frontal plane also occur. These tangential forces will cause both unfavourable bending and torsion loads on the linkage mechanism. Considering these loads cause higher (tension, pressure and shear) stresses in some of the links, they also have to be accounted for. This was done by tilting the load with $\sim 5^\circ$ and applying it at the edge of the toe, as shown in Fig.3-22c. Note that the loads F_l are applied such that no tangential forces occur in the sagittal plane (according to ISO protocol).

The loads on, and stresses in the links of the knee-ankle coupling can easily be estimated from section 3.3.2. The reader can rest assure that critical knee-ankle coupling parts are dimensioned so that they surely wont fail.

Figure 3-23 shows the final set-up of the prototype prosthesis. The prosthesis consists of a – largely- carbonfibre re-enforced toe, an aluminium forefoot, heel, and lowerleg segment and a linear heel –and plantar spring element, which are interconnected with slide bearings. The initial stiffness of the heel and the plantar spring are determined as described in sections 3.2.1 and 3.4.1 respectively. A relative weak leafspring is used to ensure lifting of the forefoot (with toe and coupler segment) during swing. The cable underneath the foot prevents the foot from collapsing during part of the period between HS and HO. Finally, the black (Delrin) block on the heel segment is intended for walking without a shoe.

The compression springs used are Fibroflex tubular spring elements. According to the manufacturer these springs have approximate linear characteristics. During self conducted measurements in which the spring elements were compressing and relaxing at a rate of 160N/s, the –approximate- linear spring characteristic had been verified. From these measurement data, a mechanical efficiency of approximately 82% was determined for the springs.

Although this mechanical efficiency is relative low when compared to coiled metal springs or carbon fibre leafsprings, using rubber compression springs has the great advantage of an easy to manipulate spring stiffness. Furthermore, rubber –compression- springs have a relative high stored energy stored per mass ratio compared to coiled metal springs, offer high strain with the low Modulus of Elasticity, and are relative easy and inexpensively available.

In the prototype foot, the plantar spring consists of one or more –serially- placed spring elements. To use these compression springs as the intended tension spring, the mechanism shown in Fig.3-23c is used. The nuts marked in Fig.3-23c function as adjustable compression stops, allowing for a progressive spring characteristic, should this be necessary.

Figure 3-23: Working prototype of described model with: a) The foot in neutral position. b) The forefoot in plantarflexion relative to the lowerleg. Compared to a), b) shows that the forefoot is in plantarflexion relative to the lowerleg with the toe in dorsalflexion relative to the forefoot. c) The flexible coupler segment enlarged. (Principle patented under number: NI1018616, PCT patent pending)

It is recognized that -in conception- the application of composites can offer an even higher stored energy per mass⁵ and possible simplification of construction by e.g. merging the lowerleg and the heel segment together into one leafspring (e.g. Fig.3-24). However, using a leaf spring as a spring element does not allow for easy manipulation of spring stiffness. Furthermore, it will probably prove difficult to correctly introduce the forces occurring during

walking to the composite leaf spring without increasing weight, *initially* complicating design and making it more costly.

In conclusion: The prototype prosthetic foot presented in this section (Fig.3-23) weighs about 520gr. and survived the static ISO testing.

Figure 3-24: Other possible embodiment of the prototype shown in Fig.3-23 where the lowerleg -and heel segments have been merged into one leafspring.

3.4.3 The Total Prototype

The previous section largely discussed the design of the prototype prosthetic foot. The total prototype however consists of the prosthetic foot and the knee-ankle coupling discussed in section 3.3.2. This section is to give a short overview of the total prototype, its' dimensions and its' adjustment possibilities.

Figure 3-25: Schematic representation of total prototype with relevant dimensions for both the foot and the knee-ankle coupling given in table.

Figure 3-25 schematically shows dimensions of the total prototype prosthesis. The basic prototype foot dimensions have been estimated according to Fig.3-9 and are based on a foot size 43. Initially variations in foot size will be compensated through elongating or shortening the toe segment. Links r_1 and r_2 are both adjustable in length. The amount of plantarflexion of link r_3 relative to link r_2 increases when link-length r_2 decreases while maintaining a vertical – start- position ($\Rightarrow$ link r_1 has to be simultaneously elongated). By employing tubular spring

elements with different hardness and lengths, the spring stiffness can be varied. In short: The link-lengths r_1 , r_2 , and the spring stiffness are adjustable in the prototype foot.

The knee-ankle coupling dimensions and spring characteristics have been discussed in section 3.3.2. In the prototypes' knee-ankle coupling the link-length r_k and the spring stiffness C_3 are also adjustable. Increasing link r_k in length causes the knee-ankle spring to tense later during a walking cycle (knee is more extended with same dorsalflexion of foot).

Figure 3-26 shows the actual total prototypes as they were used during clinical trials. The figure shows both a conventional prosthesis (upperleg support) and a PTB-type prosthesis. It is clearly seen that the knee-ankle coupling connects the upperleg directly with the prosthetic foot. In case of the PTB prosthesis this means that the prosthesis has been extended with an orthosis part.

Figure 3-26: Photograph of total prototype. a) Prototype with conventional prosthesis. b) Prototype with PTB prosthesis.

3.5 References

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3.6 Appendix

3.6.1 Gait data of normal subject used for numerical validation

(a)

(b)

(c)

(d)

Figure 3-27: Sagittal plane gait data of normal subject walking at SWS that was used for the numerical validation of the passive plantarflexing prosthetic foot. a) Right side ground reaction forces F_x and F_y as function of %WC. b) Right foot segment rotation γ from Heel Off to Toe Off as function of %WC. c) Ankle angle ψ and ankle torque M_a as function of %WC. d) Same as c) but in this graph the ankle angle is on the horizontal axis and the ankle torque is set out on the vertical axis. A positive ankle torque is plantarflexing in direction where a positive ankle rotation is dorsalflexing in direction.

3.6.2 The Four Bar Linkage

Kinematics Four Bar Linkage

Figure 3-28: Schematic representation of the four bar linkage in the prosthetic foot concept. The toe is rigidly connected to the coordinate system. a) The linkage mechanism in start position. b) With a rotation γ enforced on link r_3 .

With link length r_3 and ankle height Y determined by taking measurements (Fig.3-9), the start angle γ_s (at $H.O.$) of link r_3 to the horizontal can be determined as (Fig.3-28a):

$$\gamma_s = a \sin\left(\frac{Y}{r_3}\right) \quad 1)$$

After *choosing* the link-lengths r_2 and r_4 and the initial angles ξ and α_s of r_4 and r_2 to the horizontal and vertical respectively, link-length r_1 can be determined according to:

$$r_1 = \sqrt{x_s^2 + y_s^2} \quad \text{with:} \quad 2)$$

$$x_s = r_3 \cdot \cos(\gamma_s) - r_4 \cdot \cos(\xi) - r_2 \cdot \sin(\alpha_s)$$

$$y_s = r_3 \cdot \sin(\gamma_s) + r_4 \cdot \sin(\xi) - r_2 \cdot \cos(\alpha_s)$$

Knowing the start lengths and positions of the links relative to each other, the movements of links r_1 and r_2 can be determined as a function of the rotation γ of link r_3 relative to the start angle γ_s . The cosine rule reveals the length D as:

$$D = \sqrt{r_3^2 + r_4^2 - 2r_3 \cdot r_4 \cdot \cos(\gamma_s + \xi + \gamma)} \quad 3)$$

Again using the cosine rule, the angles δ and Δ can be derived:

$$\delta = a \cos\left(\frac{r_3^2 + D^2 - r_4^2}{2r_3 \cdot D}\right) \quad 4)$$

$$\Delta = a \cos\left(\frac{r_2^2 + D^2 - r_1^2}{2r_2 \cdot D}\right) \quad 5)$$

Determining the Instantaneous Centre of Rotation

With the positions of the different joints of the four bar linkage known, the Instantaneous Centre of Rotation (ICR) of link r_2 relative to link r_4 can be determined as it is defined to be the point of intersection of the extension lines of link r_3 and r_1 . Generally lines are described with:

$$y(x) = a \cdot x + b \quad 6)$$

Thus line r_1 is bound by the following equations (Also see Fig.3-28):

$$y_1 = a_{r1} \cdot x_1 + b_{r1} \quad 7)$$

$$y_4 = a_{r1} \cdot x_4 + b_{r1}$$

a_{r1} and b_{r1} can be solved from eq.7), thus the equation for line r_1 is defined as:

$$y(x)_{r1} = a_{r1} \cdot x + b_{r1} \quad 8)$$

$$\text{With: } a_{r1} = \frac{y_4 - y_1}{x_4 - x_1} \text{ and } b_{r1} = y_1 - a_{r1} \cdot x_1$$

The same procedure can be followed for line r_3 , thus:

$$y_3 = a_{r3} \cdot x_3 + b_{r3} \quad 9)$$

$$y_2 = a_{r3} \cdot x_2 + b_{r3}$$

a_{r3} and b_{r3} can be solved from eq.9), thus the equation for line r_3 is defined as:

$$y(x)_{r3} = a_{r3} \cdot x + b_{r3} \quad 10)$$

$$\text{With: } a_{r3} = \frac{y_2 - y_3}{x_2 - x_3} \text{ and } b_{r3} = y_3 - a_{r3} \cdot x_3$$

The coordinate x_p of the ICR can now be determined from solving the following equation:

$$y(x_p)_{r3} = y(x_p)_{r1} \Rightarrow a_{r3} \cdot x_p + b_{r3} = a_{r1} \cdot x_p + b_{r1} \quad 11)$$

$$\text{Thus: } x_p = \frac{b_{r1} - b_{r3}}{a_{r3} - a_{r1}} \quad 12)$$

Finally the coordinate y_p of the ICR can be determined by substitution of eq.12) in either eq.8) or eq.10).

Kinetics Four Bar Linkage

The previous sections described that, with the link lengths of the four bar linkage and the start angle γ_s of link r_3 to the horizontal known, the different joint positions can be determined as a function of a rotation γ of link r_3 relative to the start position. Now assume an external load F_{gr} exerted on joint 3 at an angle ε to the vertical, with the linkage mechanism in a certain position. In order for the linkage mechanism to remain in equilibrium, a torque $M_{2,3}$ is required that –in this case– is assumed to act on link r_2 and about joint 3. Because the linkage mechanism is loaded only in the joints, the line of action of F_{gr} must pass through the Instantaneous Centre of Rotation (ICR) of link r_2 and r_4 for equilibrium. Thus torque equilibrium about the joint 3 demands that:

$$M_{2,3} = F_{gr} \cdot (\cos(\varepsilon) \cdot Xp + \sin(\varepsilon) \cdot Yp) \quad 13)$$

$$\text{With: } Xp = x_p - x_3$$

$$Yp = y_3 - y_p$$

In this consideration the links are assumed to be without weight.

Figure 3-29: The four bar linkage in a certain position. An external load F_{gr} is enforced on joint 3 of the linkage mechanism at an angle ε to the vertical. A torque $M_{2,3}$ acting on link r_2 and about joint 3 is now demanded for the mechanism to remain in equilibrium.

Rotation of extra lowerleg segment with flexible joint

In the case of the plantarflexing foot, discussed in section 3.2, an extra lowerleg segment is flexibly connected to link r_2 at joint 3, as shown in Fig.3-30 (Also see Fig.3-28 and 3-29 for symbols). The figure shows the prosthetic mechanism in the start position and when link r_3 is rotated through an angle γ with an external load F_{gr} enforced on joint 3. With link r_3 rotated through an angle γ , the rotation of the lowerleg segment relative to link r_3 is built up out of the counter-clockwise rotation of link r_2 relative to r_3 (depends on the link configuration) and the clockwise rotation of the lowerleg segment relative to link r_2 , caused by the torque $M_{2,3}$ that is required for equilibrium (depends on rotational stiffness joint 3).

Figure 3-30: The plantarflexing foot prosthesis discussed in section 3.2. Basically it is the same four bar linkage as described in Figs.3-27 and 3-28 but with an extra lowerleg segment flexibly connected to link r_2 at joint 3. a) Foot in start position. b) Link r_3 rotated through an angle γ with an external force F_{gr} enforced on joint 3, at an angle ϵ to the vertical, demanding a torque $M_{2,3}$ for mechanism equilibrium.

Using the symbols set out in Fig.3-28, the rotation of link r_2 relative to r_3 is (eq.3) to eq.5):

$$\Delta\beta = \beta_s - \beta_\gamma = \delta_s + \Delta_s - \delta_\gamma - \Delta_\gamma \quad 14)$$

The rotation of the lowerleg segment relative to link r_2 caused by the torque $M_{2,3}$ is:

$$\alpha_M = \alpha_s - \alpha_\gamma \quad 15)$$

With the spring characteristic chosen such that the rotational stiffness of the lowerleg segment relative to link r_2 is such that gravity acting a mass that is connected to the lowerleg segment (COG) that rotates about joint 3 at a distance R is balanced, the rotation α_M of the lowerleg segment relative to r_2 becomes (Also see Fig.3-8):

$$\alpha_M = a \sin\left(\frac{M_{2,3}}{m \cdot g \cdot R}\right) \quad 16)$$

With: m = the mass
 R = distance of joint 3 to mass
 g = the gravitational constant

The total rotation ψ of the lowerleg segment relative to link r_3 is now defined as the difference of α_M and $\Delta\beta$:

$$\psi = \alpha_M - \Delta\beta \quad (17)$$

With $\psi > 0$, link r_3 has established a dorsalflexing rotation relative to the lowerleg segment.

With $\psi < 0$, link r_3 has established a plantarflexing rotation relative to the lowerleg segment.

Rotation of lowerleg segment with flexible link

Figure 3-31: The proposed plantarflexing foot concept discussed in section 3.4.1. Basically it is the same four bar linkage as described in Figs.3-27 and 3-28 but with link r_1 flexible. a) Foot in start position. b) Link r_3 rotated through an angle γ with an external force F_{gr} enforced on joint 3, at an angle ϵ to the vertical, demanding a torque $M_{2,3}$ for mechanism equilibrium.

In section 3.4.1 a foot concept is proposed which is basically the same as the four bar linkage described in the previous sections, only the rigid link r_1 is replaced with a flexible link. Note that in this concept the lowerleg segment and link r_2 are the same. Figure 3-30 shows such a mechanism in the start position and when link r_3 is rotated through an angle γ with an external load F_{gr} enforced on joint 3. In order for this mechanism to remain in equilibrium after a rotation γ of link r_3 , a torque $M_{2,3}$ is required that acts on link r_2 and about joint 3. Because the linkage mechanism is loaded only in the joints, the line of action of F_{gr} must pass through the Instantaneous Centre of Rotation (ICR) of link r_2 and r_4 for equilibrium (Also see Fig.3-29 and eq.13)).

However, with link r_1 flexible, it will elongate until a counteracting torque $M_{counter}$ is established that is equal to $M_{2,3}$ causing a shifting of the ICR of link r_2 and r_4 , which in turn demands a different $M_{2,3}$ for equilibrium. In other words, torque equilibrium about joint 3 also demands that:

$$M_{2,3} - M_{counter} = 0 \quad 18)$$

Revising the symbols used in Fig.3-28 learns that elongating link r_1 causes an increase in the angle Δ . With the rotational stiffness of joint 3 chosen such that gravity acting on a mass that is connected to link r_2 and rotates about joint 3 at a distance R is statically balanced, the relation for $M_{counter}$ becomes:

$$M_{counter} = m \cdot g \cdot R \cdot \sin(\Delta_\gamma - \Delta_s) \quad 19)$$

With: $m = \text{the mass}; R = \text{distance of joint 3 to mass}; g = \text{the gravitational constant}.$

By substituting eq.19) in eq.18) and solving, the link-length $r_{1\gamma}$ and the angle Δ_γ can be determined. Through solving eq.18) the rotation ψ of link r_2 (the lowerleg segment) relative to link r_3 can also be determined since:

$$\psi = \beta_s - \beta_\gamma = \delta_s + \Delta_s - \delta_\gamma - \Delta_\gamma \quad 20)$$

With $\psi > 0$, link r_3 has established a dorsalflexing rotation relative to link r_2 .

With $\psi < 0$, link r_3 has established a plantarflexing rotation relative to link r_2 .

3.6.3 Estimation initial knee-ankle coupling configuration

Figure 3-32: a) Knee-ankle coupling during standing upright. The mechanism consists of a knee lever r_k , an ankle lever r_a and a shank Sh . The linear g-spring that connects the ends of both levers is just strained with a length L_s in this position. b) Mechanism with the joint rotations occurring at 26%WC enforced on it. The lever length ratio r_k/r_a is chosen such that $L_s = L_{26\%WC}$. c) Mechanism with the joint rotations and the corresponding GRF $F_{gr_42\%}$ occurring at 42%WC enforced on it. d) The GRF $F_{gr_42\%}$ causes an external knee extension torque $M_{k_42\%WC}$ that has to be counteracted by the g-spring with a force F_t .

Section 3.3.2 discusses a knee ankle coupling mechanism consisting of a shank, an ankle lever r_a and a knee lever r_k , interconnected with hinges (Fig.3-32). A linear –gastrocnemius- tension spring (g-spring) connects the ends of both levers, spanning both hinges. In standing upright (the start position), the g-spring has a length L_s and is slightly strained (negligible, Fig.3-32a). The link length ratio of r_a and r_k is then chosen such that the g-spring also has that length L_s at approximately 26% of the WC, with relative knee and ankle rotations of 14° and 6.4° respectively enforced on the linkages (Fig.3-32b). From that point during the WC, the knee flexion torque builds as the g-spring is stretched. The (linear) g-spring stiffness C_g is now determined such that the maximum anatomic knee flexion torque of 0.27Nm/kg is established

at 42%WC with the corresponding relative anatomic knee and ankle rotations of 7.9° and 9.5° respectively, enforced on the links r_k and r_a (Fig.3-32c and 3-32d).

In the start position (Fig.3-32a), the length L_s of the g-spring and its angle γ_s relative to the shaft can be determined according to:

$$L_s = \sqrt{(r_a \cos \alpha_s - r_k)^2 + (Sh + r_a \sin(\alpha_s))^2} \quad (21)$$

$$\gamma_s = a \tan \left(\frac{r_a \cos(\alpha_s) - r_k}{Sh + r_a \sin(\alpha_s)} \right)$$

After rotating the upperleg and the foot relative to the shank about the knee and ankle joint respectively at a certain %WC, the length $L_{\%WC}$ and angle $\gamma_{\%WC}$ become (Fig.3-32b and 3-32c):

$$L_{\%WC} = \sqrt{(r_a \cos(\alpha_s + \Delta\alpha_{\%WC}) - r_k \cos(\Delta\beta_{\%WC}))^2 + (Sh + r_a \sin(\alpha_s + \Delta\alpha_{\%WC}) - r_k \sin(\Delta\beta_{\%WC}))^2} \quad (22)$$

$$\gamma_{\%WC} = a \tan \left(\frac{r_a \cos(\alpha_s + \Delta\alpha_{\%WC}) - r_k \cos(\Delta\beta_{\%WC})}{Sh + r_a \sin(\alpha_s + \Delta\alpha_{\%WC}) - r_k \sin(\Delta\beta_{\%WC})} \right)$$

By choosing either the knee lever length r_k or the ankle lever length r_a , the length of the other lever is now determined from (Fig.3-32b):

$$L_s - L_{26\%WC} = 0 \quad (23)$$

Subsequently the elongation ΔL of the g-spring with the knee -and ankle rotations occurring at 42%WC enforced on the upperleg and foot respectively, can be determined in a similar manner. Finally, with a linear g-spring and the demand that the g-spring exerts the corresponding knee flexion torque M_k at 42%WC, the g-spring stiffness C_g can be determined according to (Fig.3-32c and 3-32d):

$$C_g = \frac{F_t}{\Delta L} \quad \text{with} \quad F_t = \frac{M_{k_42\%WC}}{-r_k \sin \gamma_{42\%WC} \sin \Delta\beta_{42\%WC} + r_k \cos \gamma_{42\%WC} \cos \Delta\beta_{42\%WC}} \quad (24)$$

$$\text{With } \Delta L = L_{42\%WC} - L_s$$

By solving eq.21) to eq.24), the initial knee-ankle coupling configuration is determined.

Having determined the knee-ankle coupling configuration, the knee flexion -and ankle plantarflexion torques resulting from the g-spring stretching, can be determined as functions of enforced relative knee and ankle rotations according to:

$$\begin{aligned} M_{k_ \%WC} &= C_g \cdot (L_{\%WC} - L_s) \cdot r_k (-\sin \gamma_{\%WC} \sin \Delta\beta_{\%WC} + \cos \gamma_{\%WC} \cos \Delta\beta_{\%WC}) \\ M_{a_ \%WC} &= C_g \cdot (L_{\%WC} - L_s) \cdot r_a (-\sin \gamma_{\%WC} \sin(\alpha_s + \Delta\alpha_{\%WC}) + \cos \gamma_{\%WC} \cos(\alpha_s + \Delta\alpha_{\%WC})) \end{aligned} \quad (25)$$

4 Preliminary Prototype Testing

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4 Preliminary Prototype Testing

The previous chapter described a passive plantarflexing prosthetic foot from conception to prototype. In the course of that chapter, the mechanisms' working from Heel Off to Toe Off is numerically validated with typical normal walking at SWS as a reference. This chapter discusses the preliminary clinical testing of the prototype; the main point of interest being whether or not the prototype foot would indeed establish plantarflexion during the end of the stance phase and how gait is influenced by plantarflexion (with and without the knee-ankle coupling functionality). The discussion on the prototype testing is divided into a section where a walking model is used for prototype testing and a section that discusses preliminary clinical testing with two TT-amputees.

4.1 Preliminary testing with walking model

To minimize risks on injury and unnecessary straining of the TT-amputee, it was decided to test a 'pro-prototype' foot prior to clinical trials. For this preliminary testing a walking model has been used. This section shortly discusses the waling-model, the preliminary test set-up and its results.

4.1.1 The Homo Sapiens Walking model

As already concluded in chapter 3, it is very difficult to assess the working of a prototype prosthesis prior to clinical trials. This is mainly because it has been proven very difficult to model the human interface between prosthesis and body. Furthermore the way a TT-amputee will respond (control) to different prosthetic characteristics depends on his or her perception of the foot (and remaining capabilities) and is therefore highly subjective in nature. To get a better idea of the effects of the prototype on walking characteristics, it was decided to physically test a pro-prototype foot with some sort of human controlled walking mechanism.

With current human walking models still in a more or less experimental state, it was decided to adjust an existing human walking model already present at the Twente University such that it was usable for testing the prototypes' working. Without going into detail on the actual model creation and development, the basic working is as follows (Fig.4-1): The model makes use of a muscle driven walking segment model to validate the prototype working. The walking model is supplied with sensorial information to control the movements by three systems: propriocepsis, the visual and balance sense organs. The sensorial information is processed by a neural network such that the different segments of our so called Homo Sapiens Walking (HSW) model know what position they take in space and relative to each other.

Figure 4-1: Block diagram of the human control system.[†]

Figure 4-2 shows the actual HSW model during operation: It is seen that an orthosis is used underneath which prosthetic feet can be mounted. This enables the HSW model to walk with prosthetic feet underneath his or her sound leg. The shoe on the orthosis is equipped with a hinge joint to minimize bi-articular gastrocnemius action during walking. Furthermore the orthosis can be extended with a knee-ankle coupling as shown in the Fig4-2b (also see Fig.4-20 of Appendix).

Figure 4-2: a) The Homo Sapiens Walking model walking on a treadmill. b) The orthosis underneath which the pro-prototype foot is mounted, inclusive knee-ankle coupling.

[†]Kooij H van der, 2000, *Human Balance Control*, PhD thesis, ISBN 90-407-1963-2.

Using this HSW model had several advantages: The first being the possible human feedback on pressure spots, walking comfort. Secondly, the prototype working is evaluated under close to normal 'normal' walking conditions. Thirdly, the prototype is tested physically and not numerically thus unforeseen constructional or working hazards are immediately revealed, making it a relatively easy method for a preliminary test. Finally it can be said that, after wearing the orthosis in combination with the prosthetic foot, it is easier to interpret -and act on the comments of a prosthetic walker on a prosthesis during clinical trials.

The most obvious disadvantage is that the HSW model basically walks on stilts, causing the lowerleg to elongate by approximately 20 cm (This is accounted for in determining the COG balancing ankle stiffness). Furthermore the orthosis underneath which the prosthetic feet are mounted, causes the lowerleg to become heavier and its' center of gravity to shift distally. This last effect however has little influence during stance, and that is the part of the walking cycle that is of special interest for the prototype working. Finally it should be noted that the feedback is given by a healthy *subject(ive)*, walking with braces thus care has to be taken in translating this feedback into possible modifications in the prosthesis. On the other hand however, the healthy subject should be able to sense and describe the effects of different prototype adjustments. With that information, it is possible to get *an idea* about the effects these adjustments have on future prosthetic walkers.

The HSW model is was merely used to get an indication of the prototypes' working during future clinical trials, serving as an extra step between concept-prototype and actual conducting of clinical trials. A special point of interest was whether or not plantarflexion of the forefoot segment would be established relative to the lowerleg during the end of the stance phase with different prototype configurations. Furthermore, the effect of the knee-ankle coupling on the HSW-models' gait was assessed.

4.1.2 Test set-up and results

To assess the amount of plantarflexion established during the end of stance, goniometers were attached to the ankle -and toe hinge joints (Fig.4-3). The ankle goniometer measured the angular displacement between the forefoot -and the lowerleg segment and the toe goniometer measured the angular displacements between the toe -and the forefoot segment. Both displacements were measured relative to the segment positions for the unloaded foot. The foot used for this preliminary testing is actually a 'pro-prototype', however (although heavier) the working is exactly the same as for the prototype described in section 3.4.

During trials, the prototype prosthesis shown in Fig.4-2 and 4-3 was mounted on the left orthosis and an Otto Bock single axis foot was mounted on the right side orthosis. Subsequently the goniometer output was recorded during 30s with the HSW model standing

upright. This (static) trial served as the reference for the rotations occurring during the dynamic trials. For dynamic trials the HSW model walked on a treadmill at SWS ($\sim 1.1\text{m/s}$). During walking the HSW model was placed into a harness to prevent it from damaging in case of falling (Fig.4-2a). The angular displacements were recorded for a period of 30s., after walking comfortably for about 60s. This was done for different configurations of the prototype, with and without using the knee-ankle coupling. The relevant dimensions of the total ‘pro-prototype’ prosthesis used for these measurements and the measurement protocol are enclosed in the Appendix (section 4.4.1).

Figure 4-3: a) Schematic overview of the prosthetic foot used for preliminary testing. b) Prototype foot equipped with goniometers at the ankle joint and the toe joint.

Initially, the plantar spring stiffness was determined such that the COG of the HSW model was approximately statically balanced (section 3.4.1). The elevation of the COG relative to the prosthetic ankle joint was accounted for in determining the plantar spring stiffness (approximately 20% increase in spring stiffness). The plantar spring is built out of two serially placed compression springs with equal stiffness as described in section 3.4.2 (Fig.3-23), allowing for doubling of the spring stiffness during ambulation. If used, this was done with link r_2 (the lowerleg segment) in approximately 6° dorsalflexion relative to the link r_3 (the forefoot segment). The heel spring stiffness was determined as described in section 3.2.1. The knee-ankle coupling spring stiffness was determined as described in section 3.3.2.

During measurements, link r_2 was varied in length from 40mm to 55mm in steps of 5mm while keeping it approximately perpendicular to the ground (thus link-length r_1 was simultaneously increased). For each of the four linkage settings the plantar spring stiffness was varied from COG balancing ($\sim 15\%$ stiffer or weaker). Also plantar spring stiffness was doubled after 6° dorsalflexion of link r_2 relative to link r_3 as described in the previous paragraph.

The test results discussed in the next sections are based on experiments with two –healthy- subjects. The relevant measures of the subjects is enclosed in the Appendix (section 4.4.1)

Prototype Working

Figure 4-4 shows a typical goniometer output resulting from a measurement. The vertical axis shows the rotations of the toe segment (r_1) relative to the forefoot segment (r_3) and the rotation of the forefoot segment (r_3) relative to the lowerleg segment (r_2). Plantarflexion of the forefoot segment relative to the lowerleg segment, and dorsalflexion of the toe segment relative to the forefoot segment are negative. The maximum established plantarflexions of the forefoot (r_3) relative to the lowerleg (r_2) -averaged over a measurement session of 30s.- are also given for walking at $\sim 1.1\text{m/s}$ without knee-ankle coupling (for different link lengths r_2).

Average plantarflexion of 4.8° with $r_2=40$.

Average plantarflexion of 7.5° with $r_2=45$.

Average plantarflexion of 9.2° with $r_2=50$.

Average plantarflexion of 7.7° with $r_2=55$.

Figure 4-4: The graph shows a typical goniometer output resulting from the walking trials ($r_2=50\text{mm}$). The vertical axis shows the rotations occurring about the toe -and the ankle joint. Plantarflexion of the forefoot (r_3) relative to the lowerleg segment (r_2) and dorsalflexion of the toe segment (r_1) relative to link the forefoot segment are negative. Underneath the graph, the average maximum plantarflexion of the forefoot segment relative to the lowerleg segment are given for the different lengths of link r_2 .

It is seen that plantarflexion of the forefoot is established for every length of link r_2 . Adjusting spring stiffness relative to the approximate COG balancing stiffness (up to 15%) had negligible influence on the amount of plantarflexion established during walking. Finally, the knee-ankle

coupling also had negligible effect on the amount of plantarflexion established during the end of stance.

Figure 4-5: Two four bar linkages with different link lengths r_2 . Both are shown with the lowerleg perpendicular to the ground and with the lowerleg at an angle α to the vertical. In a) Line of action of F_{gr} does not move in front of toe. In b) Line of action of F_{gr} through tip of toe and ICR.

An averaged maximum ankle plantarflexion of $\sim 9^\circ$ was achieved with $r_2=50$; with r_2 lower -or higher, the amount of plantarflexion decreased during walking. This can be explained from considering the path of the Instantaneous Center of Rotation (ICR) of link r_2 relative to link r_4 and the line of action of the force acting on the ankle joint: Figure 4-5 shows two linkage mechanisms that are identical except for the link length r_2 . In Fig.4-5a link r_2 has a length of 50mm where the link-length of the mechanism shown in Fig.4-5b has a length of 40mm. In both mechanisms link r_2 is rotated through an angle α relative to the vertical while a force F_{gr} is applied on the ankle joint (at an angle ε to the vertical). Only considering kinematics, the mechanism with the shorter r_2 will establish a greater plantarflexion rotation of link r_3 relative to link r_2 for a certain rotation of link r_3 relative to link r_4 . However, the ICR of links r_2 and r_4 will also move further forward and into the ground for a certain rotation of link r_3 relative to link r_4 . Taking into account the force F_{gr} , equilibrium demands that the line of action of F_{gr} passes through this ICR causing the intersection of the line of action and the ground to move forward. Since the line of action of F_{gr} cannot move in front of the toe-end, eventually the foot also rotates about the tip of the toe such that the line of action of F_{gr} passes through both the ICR and the tip of the toe (Fig.4-5b). This effectively decreases the rotation of link r_3 relative to link r_4 , which results in a decrease in plantarflexion of link r_3 relative to link r_2 . Thus decreasing link length r_2 does not necessarily mean an increase in plantarflexion of link r_3 relative to link r_2 during walking.

Walking Comfort

Variation of link-length r_2 had little influence on the HSW-models perception of the prototypes' working. Although a right-left asymmetry (that increased with walking speed) was visibly present and felt during walking, it was difficult to point out the difference in sensation between the conventional foot and the prototype foot.

Spring stiffness variations (up to 15% deviation from COG balancing) had little effect on walking comfort. Sub-optimal rotational ankle joint stiffness could be easily optimized by slightly adjusting the unloaded angle of the lowerleg to the forefoot. If the ankle felt a bit weak, the forefoot was plantarflexed relative to the lowerleg where the forefoot was dorsiflexed for a stiff ankle (respectively tilting the lowerleg backward or forward for standing upright).

The knee-ankle couplings' working was as hypothesized in section 3.3.3 (Fig.3-18). Walking comfort was increased because part of the uncomfortable forces caused by the knee flexion torques occurring during walking which otherwise had to be supported by the knee (internal) and lowerleg (pressure on skin, knee hyperextension) were directly supported on the upperleg. Note that in case of the HSW models end bearing is possible where with TT amputees the majority of weight bearing is done proximal to the knee (or total contact).

4.1.3 Conclusions

During the walking trials with the HSW models plantarflexion ($\sim 9^\circ$) of the (fore)foot was established. Adjusting the length of link r_2 can optimize the maximum amount of plantarflexion established during ambulation.

The use of approximate COG balancing seems suitable as a guideline for determining the initial prosthetic ankle joint stiffness, where sub-optimal ankle stiffness (regarding walking comfort) can be optimized by slightly adjusting the unloaded angle of the lowerleg relative to the foot. Ankle joint stiffness variation had negligible influence on the amount of plantarflexion established.

The working of the knee-ankle coupling was as hypothesized in section 3.3.3, causing a decrease in knee -and lowerleg loading and therefore increasing walking comfort. The knee-ankle coupling had negligible influence on the amount of plantarflexion generated during the end of stance.

In short: The preliminary measurements indicated a correct working of the ‘pro-prototype’ and nothing indicated that the proposed mechanisms’ working should be redesigned before further developing it into a prototype suitable for clinical trials with TT-amputees (Chapter 3).

4.2 Preliminary Prototype Testing: Two Case Studies

After preliminary testing the pro-prototype with walking models, the working of the prototype discussed in Chapter 3 was assessed during clinical trials. Two TT-amputees were willing to cooperate with the conduction of these clinical trials. They will be referred to as mr.H and mr.D. This section describes the test set-up and the results of the two studies.

4.2.1 Measurement Set-up

Since –initially- the working of the prototype was of special interest, it was mounted to the shank without a shoe. A specialist correctly aligned the prosthesis was during the first session. During the following 6 ~1hour sessions the prototype foot alignment was only varied by –slightly- adjusting the link r_1 (the plantar spring link, Fig.4-6) in length; thereby tilting the shank relative to the vertical. During beginning of each session the subject was asked to walk up and down the laboratory a few times to get acquainted with the prototypes characteristics prior to measuring (for as much as this is possible in such a relative short wearing time). Subsequently, reflective markers were placed on the subjects' anatomic landmarks and the measuring session began: A total of 12 walking trials per type of prosthetic foot were conducted during which the subject walked up and down the laboratory, stepping on two subsequent Force Plates with each foot (without aiming) while the marker positions were being recorded with a six camera VICON system. Afterwards the anatomic joint kinematics and kinetics are determined (see section 2.7).

The subjects walked with two settings (A and B) of the prototype prosthetic foot, each with a different link length r_2 (Fig.4-6); to assess the knee-ankle coupling effect on walking, this was done with and without the knee-ankle coupling mounted to the total prosthesis. As already explained in the previous section, shortening of r_2 allows for more plantarflexion to be established of the forefoot relative to the lowerleg segment (per unit of toe rotation relative to the forefoot), but depends also on the subjects walking characteristics (direction GRF, angle lowerleg to the vertical).

Figure 4-6 shows the prototype foot as it was used during clinical trials. The different settings are shown in the table beside. Furthermore the estimated required spring stiffness' for mr.H as well as for mr.D are given. The ankle joint -and knee-ankle coupling spring stiffness (C_1 , C_2 and C_3 respectively) have been determined as described in sections 3.2.1; 3.4.1, and 3.3.2 respectively, with the dimensions of the prototype foot according to setting A but with $r_2=55\text{mm}$.

The main point of interest was the prototypes' working under walking conditions; the main question being whether or not plantarflexion of the (fore)foot would be established during the

end of affected stance (while displaying a more natural power characteristic) but also if the knee-ankle coupling would function as hypothesized. Furthermore, the effects of the prototypes' characteristics on gait and -more importantly- on the amputees' perception of walking were of interest.

The next sections subsequently discuss the results for walking with the prosthesis currently in use, and for walking with the prototype foot in the different settings (with and without the knee-ankle coupling). Data resulting from walking with current prosthesis will be compared with average normal gait (see Chapter 2) whereas data resulting from walking with the prototype prosthesis will be compared to walking with the current prosthesis. The reader is referred to the chapter 2 for the sign conventions (Fig.2-2 and Fig.2-5).

(a)

	<i>Dimensions prototype</i>		
Setting A	As Fig.4-25 with $r_2=56.5\text{mm}$ and $r_k\sim 30\text{mm}$		
Setting B	As Fig.4-25 with $r_2=53.5\text{mm}$ and $r_k\sim 30\text{mm}$		
	C_1 (N/mm)	C_2 (N/mm)	C_3 (N/mm)
Mr. H	90	265	180
Mr. D	110	325	220

(b)

Figure 4-6: a) Prototype prosthetic foot in neutral position. b) Different dimensions and different settings of the prototype as it was used during clinical trials (Also see Fig.3-25).

4.2.2 Mr.H and results of walking with prototype

Mr.H (23-12-1934) is a right-sided transtibial amputee. Cause of the amputation was a trauma. At first Mr.H had been fitted with a Patellar Tendon Bearing socket. However, because of a worn out knee on the prosthetic side, he now uses a conventional socket (support on upperleg with thigh cuff, Fig.3-26). Mr. H also wears an orthopedic shoe on his sound leg. Nevertheless Mr.H is a very active walker with no real complaints about his current prosthesis except for the fact that by using the conventional socket one can see the prosthesis. Currently he is using a single axis foot. The relevant measures of Mr.H are enclosed in the Appendix (section 4.4.2).

Walking with current prosthesis

Temporal Trends (Table 4-1) and Kinematics (Fig.4-7)

<i>Measurement</i>	<i>Current prosthesis</i>
<i>Velocity</i> (m/s)	1.21
<i>Affected stance time</i> (%WC)	62.6
<i>Sound stance time</i> (%WC)	64.9
<i>Affected double support</i> (%WC)	11.4
<i>Sound double support</i> (%WC)	16.1
<i>Affected step length</i> (m)	0.72
<i>Sound step length</i> (m)	0.73

Table 4-1: temporal trend and kinematics for walking with the current prosthesis.

Figure 4-7: Joint angular rotations established during walking with current prosthesis. RHIP, RKNE, and RANK are the Right hip, Right knee, and Right ankle joints respectively. LHIP, LKNE, and LANK are the Left hip, Left knee and Left ankle joints respectively.

Considering the temporal trends of mr.H, it is seen that the affected and sound step-lengths are almost symmetric. Affected double support and stance time however are both shorter than sound double support and stance time.

Considering affected and sound limb joint angles, it is seen that the affected ankle joint rotation shows little variation (except for some plantarflexion during weight acceptance) and lacks plantarflexion during the end of affected stance. This is as expected with a –more or less- rigid ankle joint function. The sound ankle also displays a decreased plantarflexion during late stance, which might be the result of the orthopedic shoe he wears (because of arthritis).

Affected and sound hip flexion angles established during affected and sound weight acceptance respectively, are both higher than average normal (~74% and 60% increase respectively) where the established affected hip flexion is slightly (8%) higher than the sound hip flexion. Also, affected and sound knee remain more in extension during affected and sound Toe Off respectively when compared to average normal (affected knee flexion 58% -and sound knee 67% of normal). This is typical for TT-amputee gait (Chapter 2).

Kinetics (Fig.4-8)

The peak affected and sound hip extension torques occurring during weight acceptance are both substantially higher than normal (~156% and 127% increase respectively); the peak sound extension torque is 88% of the peak affected hip extension torque. This can also be found in the power graph where more mechanical energy is generated about the affected hip joint, especially during early stance. This concurs with the increased hip flexion angles during weight acceptance. It is also seen that the sound hip extension torque starts quite abruptly at sound Heel Strike. The Net amount of Mechanical Energy (NME) generated about the affected and sound hip joints are 46J and 40J respectively. The NME absorbed about the affected and sound hip joints are 0J and 3J respectively.

The peak affected knee extension torque during affected weight acceptance is approximately 40% of normal. During late stance the affected knee torque remains flexor in direction of which the peak lies within normal range. On the sound side mr.H displays increased knee extension and flexion torques compared to normal (25% to 50% increase). This concurs with the forward trunk-lean and the tendency of an amputee to use the sound side (increasing sound side loading) to augment forward progression of the center of gravity. The NME generated about the affected and sound knee joints are 4J and 8J respectively. The NME absorbed about the affected and sound knee joints are 13J and 28J respectively.

The peak affected ankle joint torque is lower than the sound ankle joint torque (which is close to normal). From the power graphs it is seen that the affected ankle joint acts mainly as an energy absorber, almost totally lacking the typical power-burst that is needed to establish normal push-off. Although the sound ankle joint displays the power characteristics that are typical for normals (inclusive power burst during the end of sound stance), it is also a net

energy absorber: The NME generated about the affected and sound ankle joints are 5J and 16J respectively. The NME absorbed about the affected and sound ankle joints are 10J and 19J respectively.

Figure 4-8: Joint kinetics displayed during walking with the current prosthesis. RHIP, RKNE, and RANK are the Right hip, Right knee, and Right ankle joints respectively. LHIP, LKNE, and LANK are the Left hip, Left knee and Left ankle joints respectively. RFy and RFx are the vertical and horizontal (in the plane of progression) Ground Reaction Force components established by the right leg respectively. LFy and LFx are the vertical and horizontal (in the plane of progression) Ground Reaction Force components established by the left leg respectively. a) Joint torques. b) Joint powers. c) Ground Reaction Forces

Finally it is seen that the first vertical sound Ground Reaction Force peak (during weight acceptance) is substantially (~20%) higher than average normal. The second sound side and

both affected side vertical GRF peaks are lower than average normal (~92%-95% of normal). Also the second horizontal (propulsive) sound GRF is slightly (~5%) higher than normal. All in all, mr.H displayed typical kinematic and kinetics of a TT-amputee.

Walking with prototype foot in setting A

Temporal Trends (Table 4-2) and Kinematics (Fig.4-9)

Measurement	Prosthetic foot type		
	Current	Setting A	Setting A with KA coupling
Velocity (m/s)	1.21	1.15	1.15
Affected stance time (%WC)	62.6	65.8	62.1
Sound stance time (%WC)	64.9	65.8	66.2
Affected double support (%WC)	11.4	12.2	11
Sound double support (%WC)	16.1	19.3	17.3
Affected step length (m)	0.72	0.69	0.72
Sound step length (m)	0.73	0.74	0.71

Table 4-2: temporal trend and kinematics for walking with the prototype foot in setting A.

Figure 4-9: Joint angular rotations established during walking with the prototype in setting A. RHIP, RKNE, and RANK are the Right hip, Right knee, and Right ankle joints respectively. LHIP, LKNE, and LANK are the Left hip, Left knee and Left ankle joints respectively.

In walking with the prototype in setting A, it is seen that walking velocity decreased slightly (~3%). This can be the result of a lack of trust in the prototype because of the relative short wearing period. However, sound step-length increased relative to affected step-length, but also slightly relative to the sound step length when walking with the current prosthesis. Sound—and affected stance times are both longer and symmetric. Finally, affected and sound double support times increased but also did their asymmetry (can be because of the relative short time

to get acquainted with the different prototypes' plantarflexing action during weight acceptance).

Considering affected and sound limb joint angles, it is seen that the prototype foot established about 18° plantarflexion under loading, which is more than the plantarflexion established on the sound side. Compared to walking with the current prosthesis, both the affected and sound hip joint are more in extension during affected and sound Toe Off respectively (5° and 6° respectively). The sound hip joint is slightly (~3°) more in extension during sound Toe Off compared to the affected hip angle during affected Toe Off (which concurs with the longer sound double support time). During weight acceptance both the affected and sound hip are also more in extension when compared to the current prosthesis (~17% less peak flexion). Finally, it is seen that the peak sound knee flexion during stance decreases to 60% of walking with the current prosthesis.

Temporal trends and kinematics are not influenced much when the knee-ankle coupling was mounted to this prosthetic setup. Affected Toe Off is established slightly earlier during a Walking Cycle, which causes the decreased affected stance and double support times. This is probably caused by the fact that the knee-ankle coupling aids in establishing Toe Off during a large part of the affected stance. Mind you that this decrease in affected stance times can also because these are results from the first session (of six) during which the subject had been walking with the coupling and was -most likely- not entirely accustomed with the mechanisms characteristics. Biomechanical joint data of walking with knee-ankle coupling are enclosed in the Appendix (section 4.4.3) and table 4-2.

Kinetics (Fig.4-10)

When comparing affected and sound hip joint extension torques during weight acceptance, it is seen that the affected side remains higher than the sound side and both peaks are approximately the same as when the subject was walking with the current prosthesis (although the sound hip extension torque displays an overall decrease). It is again seen that the sound hip extension torque starts quite abruptly at sound Heel Strike. The peak sound hip flexion torque during late stance increased (44%). The hip joint power characteristics are also approximately the same as when walking with the current prosthesis. The NME generated about the affected and sound hip joints are 55J and 37J respectively. The NME absorbed about the affected and sound hip joints are 1J and 2J respectively.

Peak affected knee joint torques have increased (~100% more extension), where sound knee joint torques have decreased (55% less extension and 22% less flexion) from ~10% into stance to end of stance. Making the sound and affected knee joint torques reasonably normal. The NME generated about the affected and sound knee joints are 5J and 7J respectively. The NME absorbed about the affected and sound knee joints are 15J and 20J respectively.

Figure 4-10: Joint kinetics displayed during walking with the prototype in setting A. RHIP, RKNE, and RANK are the Right hip, Right knee, and Right ankle joints respectively. LHIP, LKNE, and LANK are the Left hip, Left knee and Left ankle joints respectively. RFy and RFx are the vertical and horizontal (in the plane of progression) Ground Reaction Force components established by the right leg respectively. LFy and LFx are the vertical and horizontal (in the plane of progression) Ground Reaction Force components established by the left leg respectively. a) Joint torques. b) Joint powers. c) Ground Reaction Forces

Peak affected ankle plantarflexing torque has increased where the sound ankle plantarflexing torque has decreased compared to walking with the current prosthesis. It is now even seen that peak affected plantarflexing ankle joint torque is higher than peak sound plantarflexing ankle torque. The affected dorsalflexing torque just after heel strike has increased in magnitude and in duration. The affected and sound ankle joint now both display a power burst during the end

of stance. However it is again seen that both still are mainly energy absorbers: The NME generated about the affected and sound ankle joints are 28J and 14J respectively. The NME absorbed about the affected and sound ankle joints are 32J and 21J respectively.

Comparing the GRF shows that the first sound horizontal (40% decrease) and vertical (22% decrease) GRF peaks (during sound weight acceptance) are substantially lower compared to walking with the current prosthesis. Also the second horizontal GRF peak shows a 16% decrease. The first affected horizontal GRF peak is 25% lower while the second peak is 14% lower. Finally the second affected vertical GRF peak was increased by 6%.

Peak joint torques, powers, overall joint kinetics characteristics and the Ground Reaction Force are not influenced much by the knee-ankle coupling other than that affected knee extension torque just after Heel Strike is increased (~20%). This results in a higher affected power absorption peak during weight acceptance. Biomechanical joint data of walking with knee-ankle coupling are enclosed in the Appendix (section 4.4.3) and table 4-2.

Walking with prototype foot in setting B

Temporal Trends (Table 4-3) and Kinematics (Fig.4-11)

Comparing the temporal trends of walking with the prototype in setting B to walking with the current prosthesis displays an overall improved symmetry while walking speed is increased compared to walking with setting A. Furthermore affected double support time is increased while sound double support time is decreased (compared to walking with setting A); improving gait symmetry. Compared to walking with the current prosthesis, affected step-length and (double) stance-time have increased. These data indicate that mr.H is now more accustomed to the prototypes' walking characteristics.

<i>Measurement</i>	<i>Prosthetic foot type</i>		
	<i>Current</i>	<i>Setting B</i>	<i>Setting B with KA coupling</i>
<i>Velocity (m/s)</i>	1.21	1.18	1.17
<i>Affected stance time (%WC)</i>	62.6	64.2	64.4
<i>Sound stance time (%WC)</i>	64.9	65.1	64.4
<i>Affected double support time (%WC)</i>	11.4	12.9	12.1
<i>Sound double support time (%WC)</i>	16.1	16.5	16.7
<i>Affected step length (m)</i>	0.72	0.71	0.71
<i>Sound step length (m)</i>	0.73	0.71	0.73

Table 4-3: temporal trend and kinematics for walking with the prototype foot in setting B.

Comparing affected and sound limb joint angles for walking with setting B shows and affected ankle joint plantarflexion under loading of approximately 17° . Compared to walking with the current prosthesis, both the affected and sound hip joint during affected and sound weight acceptance respectively, are slightly more in extension ($2-3^\circ$) and reasonable symmetric. Also the peak affected knee flexion during affected stance increased 6° (approaching normal). Finally, it should be noted that due to technical problem, the prototypes' forefoot did not flex back to neutral after loading; this is why the forefoot remains in a plantarflexing position during swing. However, this could not have effected the stance phase since little force was needed to push the foot back to the unloaded position.

Figure 4-11: Joint angular rotations established during walking with the prototype in setting B. RHIP, RKNE, and RANK are the Right hip, Right knee, and Right ankle joints respectively. LHIP, LKNE, and LANK are the Left hip, Left knee and Left ankle joints respectively.

Temporal trends and kinematics are not influenced much when the knee-ankle coupling is mounted to this prosthetic foot set-up, although stance time symmetry has slightly improved. Compared to walking with the knee-ankle coupling in combination with setting A, this session displays more symmetry. Biomechanical joint data of walking with the knee-ankle coupling are enclosed in the Appendix (section 4.4.3) and table 4-3.

Kinetics (Fig.4-12)

When comparing affected and sound hip joint extension torques just after Heel Strike, it is seen that the affected side is higher than the sound side but both sound -and affected side are approximately the same as when the subject was walking with the current prosthesis (sound side peak extension torque is slightly higher but displays an overall decrease). During late affected and sound stance/ early swing the affected and sound hip flexion torques increased ($\sim 20\%$). Also it is seen that the peak affected hip flexion torque occurs earlier during the WC. The sound hip extension torque starts quite abruptly just after sound Heel Strike. The hip joint power characteristics are approximately the same as when walking with the current prosthesis. The NME generated about the affected and sound hip joints are 53J and 35J respectively. The NME absorbed about the affected and sound hip joints are 4J and 2J respectively.

Peak affected knee joint torques have increased (~200% more extension and 50% more flexion), where sound knee joint torques have decreased (60% less extension and 10% less flexion) from ~10% into stance to end of stance. The NME generated about the affected and sound knee joints are 5J and 7J respectively. The NME absorbed about the affected and sound knee joints are 18J and 21J respectively.

Figure 4-12: Joint kinetics displayed during walking with the prototype in setting B. RHIP, RKNE, and RANK are the Right hip, Right knee, and Right ankle joints respectively. LHIP, LKNE, and LANK are the Left hip, Left knee and Left ankle joints respectively. RFy and RFx are the vertical and horizontal (in the plane of progression) Ground Reaction Force components established by the right leg respectively. LFy and LFx are the vertical and horizontal (in the plane of progression) Ground Reaction Force components established by the left leg respectively. a) Joint torques. b) Joint powers. c) Ground Reaction Forces.

The peak affected ankle joint torque is approximately the same as for walking with setting A (increased compared to current prosthesis) where the peak sound ankle joint torque is now approximately equal to walking with the current prosthesis (and the affected ankle). The affected dorsalflexing torque just after heel strike has again increased in magnitude and in duration. The affected and sound ankle joint now *both* display the typical a power burst during the end of stance. The NME generated about the affected and sound ankle joints are 33J and 16J respectively. The NME absorbed about the affected and sound ankle joints are 26J and 23J respectively.

Comparing the GRF shows that the first sound horizontal (60% decrease) and vertical (14% decrease) GRF peaks (during sound weight acceptance) are substantially lower compared to walking with the current prosthesis. Also the second –sound- horizontal GRF peak shows 8% decrease. The first affected horizontal GRF peak is 28% lower while the second peak is 17% lower. Finally, the first affected vertical GRF peak increased by 6% and the second with 11%. Joint kinetics are not influenced much by the knee-ankle coupling other than that the first (weight-acceptance) sound side peak in GRF increased to –almost (92% of walking with current)- the value of walking with the current prosthesis. Also the second vertical affected side GRF peak decreased to the value of walking with the current prosthesis. Affected knee extension torque after weight acceptance is slightly increased (~10%) which also resulted in a higher affected power absorption peak after weight acceptance. Biomechanical joint data of walking with knee-ankle coupling are enclosed in the Appendix (section 4.4.3) and table 4-3.

Mr.H perception

During the first sessions Mr.H felt more insecure and less stable while walking with the prototype foot. This feeling decreased as the number of sessions increased, indicating that a period is needed for the subject to get accustomed to the prototype characteristics. Walking with the prototype did not at any time during measurements cause more pain or walking discomfort compared to walking with his current prosthesis.

With respect to walking with the prototype foot in both settings, Mr.H felt a clear lifting or push off sensation during the end of affected stance. He felt as though the prototype made unwinding of the foot during the end of stance easier, which resulted in a less strenuous walking pattern. Also walking with the foot in setting B was easier than with setting A. Since setting B was used during the final measurements and biomechanical analysis of the affected ankle joint showed no significant differences with setting A, this improvement is most likely the result of the user getting more acquainted to the prototype working rather than the result of the different prototype settings. Although mr.H rated the push off sensation as good (from good/ same / worse) compared to his current prosthesis, he also remarked this was a difficult question to answer because of the relative short wearing time of the prototype.

When walking with the prototype foot in combination with the knee ankle coupling, biomechanical data reveal little difference in joint kinetics and kinematics; although affected knee extension torque during affected weight acceptance increased. However, mr.H almost immediately remarked (without being asked) that walking became less strenuous, laborious and more comfortable. He felt as though unwinding became easier. Finally he felt more stable while walking with the coupling, probably due to the fact that he had more control over affected push off. Adjusting the lever-length (plus or minus 3-5mm) at the knee joint (r_k , Fig.4-6) had little or no influence on the walking sensation.

4.2.3 Mr.D and results of walking with prototype

Mr.D (24-12-1930) is a left-sided transtibial amputee. Cause of the amputation was a trauma. Mr.D is fitted with a Patellar Tendon Bearing suction socket. Mr.D is an active walker and currently uses a single axis foot. The relevant measures of mr.D are enclosed in the Appendix (section 4.4.2).

Walking with current prosthesis

Temporal Trends (Table 4-4) and Kinematics (Fig.4-13)

Considering the temporal trends of mr.D, it is seen that the affected stance time, double support time and step-length all have lower values than on the sound side.

<i>Measurement</i>	<i>Current prosthesis</i>
<i>Velocity (m/s)</i>	1.02
<i>Affected stance time (%WC)</i>	63.4
<i>Sound stance time (%WC)</i>	66.3
<i>Affected double support (%WC)</i>	11.9
<i>Sound double support (%WC)</i>	17.9
<i>Affected step length (m)</i>	0.63
<i>Sound step length (m)</i>	0.68

Table 4-4: temporal trend and kinematics for walking with the current prosthetic foot.

When considering affected and sound limb joint angles, it is seen that the affected ankle joint rotation shows little variation (except for some plantarflexion during weight acceptance) and lacks plantarflexion during the end of affected stance. This is as expected with a –more or less– rigid ankle joint function.

Affected (41% increase) and sound (60% increase) hip angles established during affected and sound weight acceptance respectively, are both higher than normal. Also, the affected hip angle is more in extension (4° more) where the sound hip is more in flexion ($\sim 25\%$ more) during affected Toe Off (compared to sound and affected hip angles during sound side Toe Off respectively), indicating a forward trunk lean during affected Toe Off to augment forward progression of the trunk.

Figure 4-13: Joint angular rotations established during walking with the current prosthetic foot. RHIP, RKNE, and RANK are the Right hip, Right knee, and Right ankle joints respectively. LHIP, LKNE, and LANK are the Left hip, Left knee and Left ankle joints respectively.

Affected and sound knee remain more in extension during affected and sound Toe Off respectively (affected knee flexion 57% -and sound knee 62% of normal). Also the affected knee remains more in extension throughout affected stance compared to the sound knee during sound stance. These gait characteristics are typical for TT-amputee gait (Chapter 2).

Kinetics (Fig.4-14)

During weight acceptance the peak affected and sound hip extension torques are both substantially higher than average normal (51% and 71% respectively). The sound side hip extension torque during weight acceptance is higher than the affected side, which concurs with the simultaneously occurring increased sound side hip flexion angle (forward trunk lean). The NME generated about the affected and sound hip joints are 36J and 34J respectively. The NME absorbed about the affected and sound hip joints are 3J and 5J respectively.

The peak affected knee extension torque during affected weight acceptance is approximately 50% lower than normal. During late stance the affected knee torque remains extensor in direction of which the peak lies within normal range. On the sound side mr.D displays increased knee extension torque compared to normal ($\sim 15\%$). This concurs with the forward trunk-lean and the tendency of the amputee to use the sound side (increasing sound side loading) to augment forward progression of the center of gravity. The NME generated about the affected and sound knee joints are 2J and 6J respectively. The NME absorbed about the affected and sound knee joints are 16J and 24J respectively.

Figure 4-14: Joint kinetics displayed during walking with the current prosthetic foot. RHIP, RKNE, and RANK are the Right hip, Right knee, and Right ankle joints respectively. LHIP, LKNE, and LANK are the Left hip, Left knee and Left ankle joints respectively. RFy and RFx are the vertical and horizontal (in the plane of progression) Ground Reaction Force components established by the right leg respectively. LFy and LFx are the vertical and horizontal (in the plane of progression) Ground Reaction Force components established by the left leg respectively. a) Joint torques. b) Joint powers. c) Ground Reaction Forces.

The peak affected ankle joint torque is lower than the sound ankle joint torque (which lies within normal range). From the power graphs it is seen that the affected ankle joint acts mainly as an energy absorber, displaying a significantly reduced power-burst prior to Toe Off. The sound ankle joint displays the power characteristics that are typical for normals (inclusive power burst during the end of sound stance). The NME generated about the affected and sound

ankle joints are 6J and 25J respectively. The NME absorbed about the affected and sound ankle joints are 15J and 20J respectively.

Finally, both sound side horizontal GRF peaks are ~65% of average normal. The first affected horizontal GRF peak is ~35% of normal, the second peak ~55%. The second sound side and both affected side vertical GRF peaks are lower than normal (~89% to 81% of normal). All in all mr.D displayed the similar typical TT-amputee gait characteristics as mr.H only with more asymmetry and lower peaks.

Walking with prototype foot in setting A

Temporal Trends (Table 4-5) and Kinematics (Fig.4-15)

<i>Measurement</i>	<i>Type Prosthetic Foot</i>		
	<i>Current</i>	<i>Setting A</i>	<i>Setting A with KA coupling</i>
<i>Velocity (m/s)</i>	1.02	1.01	0.99
<i>Affected stance time (%WC)</i>	63.4	63.6	62.4
<i>Sound stance time (%WC)</i>	66.3	64.7	65.0
<i>Affected double support (%WC)</i>	11.9	8	8
<i>Sound double support (%WC)</i>	17.9	20.3	19
<i>Affected step length (m)</i>	0.63	0.64	0.64
<i>Sound step length (m)</i>	0.68	0.7	0.7

Table 4-5: temporal trend and kinematics for walking with the prototype foot in setting A.

Figure 4-15: Joint angular rotations established during walking with the prototype in setting A. RHIP, RKNE, and RANK are the Right hip, Right knee, and Right ankle joints respectively. LHIP, LKNE, and LANK are the Left hip, Left knee and Left ankle joints respectively.

In walking with the prototype in setting A, it is seen that affected stance time increased (marginally) while sound stance time decreased, improving symmetry. However, affected double support time decreased while sound double support time increased, increasing double

support asymmetry. This is most likely the result of a lack of trust in the prototype because of the relative short wearing period (this being one of the first sessions during which the subject walked with the prototype). However, despite of a possible lack of trust in the prototype, sound step-length and affected stance time increased (marginally).

Considering affected and sound limb joint angles, it is seen that the prototype foot established about 15° plantarflexion under loading. During affected weight acceptance the affected hip is slightly more in extension when compared to the current prosthesis (~7%). Also, the sound hip joint is slightly more in extension during sound Toe Off (~2°).

There is little influence of the knee-ankle coupling on the temporal trends: Affected stance time decreases while sound stance time increases. Only the sound double support time displays a slight decrease. Joint kinematics are also not influenced much by the use of the knee-ankle coupling. However, the sound hip joint is more in extension during stance, and is now almost equal to the affected hip rotation. Also the affected knee remains in a more flexion position during stance. Biomechanical data of walking with the knee-ankle coupling are enclosed in the Appendix (section 4.4.3) and table 4-5.

Kinetics (Fig.4-16)

When comparing affected and sound hip joint extension torques during weight acceptance, it is seen that the affected side remains higher than the sound side. The peak sound hip extension torque during weight acceptance remains approximately the same whereas the affected hip extension torque reduces with 13%. The peak sound hip flexion torque during late sound stance increased with 65%. This is also seen in the hip joint power characteristics: The NME generated about the affected and sound hip joints are 29J and 32J respectively. The NME absorbed about the affected and sound hip joints are 1J and 7J respectively.

The peak affected knee extension torque after weight acceptance has decreased with 40%, where the peak affected knee flexion torque during affected stance has increased from 0 to 22.5Nm, which is approximately normal. Peak sound knee extension during weight acceptance has decreased to 14% of walking with the current prosthesis where the peak sound knee flexion torque during stance has increased from 7 -to 29Nm (~average normal). The NME generated about the affected and sound knee joints are 1J and 3J respectively. The NME absorbed about the affected and sound knee joints are 12J and 17J respectively.

Peak affected ankle plantarflexing torque has increased where the sound ankle plantarflexing torque has decreased compared to walking with the current prosthesis; both peaks are approximately the same. The affected dorsalflexing torque just after heel strike has increased in magnitude and in duration. The affected and sound ankle joint now both display a power burst during the end of stance. The NME generated about the affected and sound ankle joints are 24J and 21J respectively. The NME absorbed about the affected and sound ankle joints are 10J and 17J respectively.

Comparing the GRF shows that the second affected horizontal (propulsive) GRF peak has increased with 35%.

Figure 4-16: Joint kinetics displayed during walking with the prototype in setting A. RHIP, RKNE, and RANK are the Right hip, Right knee, and Right ankle joints respectively. LHIP, LKNE, and LANK are the Left hip, Left knee and Left ankle joints respectively. RFy and RFx are the vertical and horizontal (in the plane of progression) Ground Reaction Force components established by the right leg respectively. LFy and LFx are the vertical and horizontal (in the plane of progression) Ground Reaction Force components established by the left leg respectively. a) Joint torques. b) Joint powers. c) Ground Reaction Forces

Joint kinetics and the GRF are not influenced much by the knee-ankle coupling other than that affected knee torque is now mainly flexor in direction during stance and larger in magnitude.

This resulted in a higher affected knee power peak during stance. Biomechanical joint data of walking with knee-ankle coupling are enclosed in the Appendix (section 4.4.3) and table 4-5.

Walking with prototype foot in setting B

Temporal Trends (Table 4-6) and Kinematics (Fig 4-17)

<i>Measurement</i>	<i>Type Prosthetic Foot</i>		
	<i>Current</i>	<i>Setting B</i>	<i>Setting B with KA coupling</i>
<i>Velocity (m/s)</i>	1.02	1.07	1.01
<i>Affected stance time (%WC)</i>	63.4	64.2	65.2
<i>Sound stance time (%WC)</i>	66.3	64.6	64.2
<i>Affected double stance time (%WC)</i>	11.9	10	10.3
<i>Sound double stance time (%WC)</i>	17.9	19	19.1
<i>Affected step length (m)</i>	0.63	0.65	0.64
<i>Sound step length (m)</i>	0.68	0.71	0.73

Table 4-6: temporal trend and kinematics for walking with the prototype foot in setting B.

Figure 4-17: Joint angular rotations established during walking with the prototype in setting B. RHIP, RKNE, and RANK are the Right hip, Right knee, and Right ankle joints respectively. LHIP, LKNE, and LANK are the Left hip, Left knee and Left ankle joints respectively.

Comparing the temporal trends of walking with the prototype in setting B to walking with the current prosthesis shows an increased walking velocity (5%). Both affected and sound step length increased (5% and 3% respectively). Affected stance time increased while sound stance time decreased improving stance symmetry. Affected double support increased while sound double support decreased relative to walking with setting A. However compared to walking with the current prosthesis affected double support time remains lower while sound stance time remains higher. These results suggest that the subject has become more acquainted with the prototype characteristics.

Considering affected and sound limb joint angles, it is seen that the prototype foot established about 18° plantarflexion under loading, which is approximately equal to the sound side.

Compared to walking with the current prosthesis, both affected and sound hip joint angles rotations are more in extension during affected and sound Toe Off respectively (4° and 2° respectively). When compared to walking with the current prosthesis, the affected hip is more in extension during affected weight acceptance (~25% less flexion).

The influence of the knee-ankle coupling on the temporal trends and joint kinematics is negligible, although affected stance time has increased and walking velocity has decreased. Biomechanical joint data of walking with knee-ankle coupling are enclosed in the Appendix (section 4.4.3) and table 4-6.

Kinetics (Fig 4-18)

Compared to walking with the current prosthesis, the peak affected and sound hip joint extension torques during weight acceptance decreased with 30% and 12% respectively. In this case the peak sound hip extension torque is 35% higher than affected hip extension torque. The peak sound hip flexion torque during late sound stance increased with 70%. This is also seen in the hip joint power characteristics: The NME generated about the affected and sound hip joints are 27J and 22J respectively. The NME absorbed about the affected and sound hip joints are 3J and 8J respectively.

The peak affected knee extension torque after affected weight acceptance has slightly increased with ~13%. Peak sound knee extension after weight acceptance has decreased to 42% of walking with the current prosthesis where the peak sound knee flexion torque during stance has increased with 250% (to approximately average normal). The NME generated about the affected and sound knee joints are 2J and 3J respectively. The NME absorbed about the affected and sound knee joints are 15J.

Peak affected ankle plantarflexing torque has increased where the sound ankle plantarflexing torque has decreased compared to walking with the current prosthesis; both peaks are approximately the same. The affected and sound ankle joint now both display a power burst during the end of stance. The NME generated about the affected and sound ankle joints are 30J and 29J respectively. The NME absorbed about the affected and sound ankle joints are 14J and 24J respectively.

Comparing the GRF, it is again seen that the second affected horizontal (propulsive) GRF peak has increased (30%).

Joint kinetics and the GRF are not influenced much by the knee-ankle coupling other than that affected knee torque is now mainly flexor in direction during stance and larger in magnitude. Yet the affected knee power absorption during –the end of- affected stance has reduced. Biomechanical joint data of walking with knee-ankle coupling are enclosed in the Appendix (section 4.4.3) and table 4-6.

Figure 4-18: Joint kinetics displayed during walking with the prototype in setting B. RHIP, RKNE, and RANK are the Right hip, Right knee, and Right ankle joints respectively. LHIP, LKNE, and LANK are the Left hip, Left knee and Left ankle joints respectively. RFy and RFx are the vertical and horizontal (in the plane of progression) Ground Reaction Force components established by the right leg respectively. LFy and LFx are the vertical and horizontal (in the plane of progression) Ground Reaction Force components established by the left leg respectively. a) Joint torques. b) Joint powers. c) Ground Reaction Forces.

Mr.D perception

As was also the case for mr.H, mr.D felt more insecure and less stable while walking with the prototype foot. Also in this case this feeling decreased as the number of sessions increased, indicating that a period is needed for the subject to get accustomed to the prototype

characteristics. Walking with the prototype did not at any time during the measurements cause more pain or walking discomfort compared to walking with his current prosthesis.

With respect to walking with the prototype foot in both settings, Mr.D felt less resistance during push off and also a better push off compared to walking with his current prosthesis. Mr. D did not really feel any difference in walking with the different prototype settings. Because of the relative short wearing time of the prototype it was not possible to determine whether or not the new prosthetic characteristics –that were clearly felt- were an improvement compared to walking with his current prosthesis.

When walking with the prototype foot in combination with the knee-ankle coupling, the fact that an orthotic extension was required for the knee-ankle coupling was an issue for mr.D from a cosmetic and easy donning and doffing point of view. He did not see the need for wearing that orthotic extension when he had done without for many years now. During the first sessions with the knee-ankle coupling mr.D did not feel any difference compared to walking without the coupling. However, gait data reveals that affected knee kinematics during walking with the knee-ankle coupling altered significantly (to mainly flexor in direction). The knee-ankle coupling action was confirmed by the video-recordings that clearly showed compression of the knee-ankle coupling spring during ambulation. It was not until the last sessions in walking with setting B and the knee-ankle coupling that mr.D became very enthusiastic about the knee-ankle coupling. He remarked (without asking) that walking became less strenuous and walking with the knee-ankle coupling walks wonderful. After analyzing this session, gait characteristics were found that were similar to the characteristics already found. This suggests that mr.D needed time to get used to the knee-ankle coupling characteristics. Finally, as was also the case for mr.H: Adjusting the lever-length (plus or minus 3-5mm) at the knee joint (r_k , Fig.4-6) had little or no influence on the walking sensation of mr.D.

4.3 Resume

In chapter 3 a passive prosthetic foot mechanism that achieves leg-elongation during the end of the stance phase by means of plantarflexion of the foot has been proposed. The energy needed to establish the plantarflexion is borrowed from and given back to the total prosthetic anatomic system in which human walking is considered to be a ballistic process. In combining a four bar linkage with an approximate center of gravity balancing ankle joint, the plantarflexing foot aims to re-introduce the typical power-burst during the end of prosthetic stance while supporting the trunk at the right height (demanding a plantarflexing rotation of the foot). It is argued that this allows for enough energy to be transferred from the affected to the sound side in a more natural manner, and allows for a more natural progression and –symmetric- support of the trunk, thereby improving prosthetic walking comfort and reducing the chance on long term secondary complaints such as lower back pain and early degenerative arthritis.

Aiming to further improve on walking comfort (decrease of residual limb loading and a more natural force transmission between joints), the lowerleg prosthesis allows for an optional coupling of the anatomic knee and prosthetic ankle joint during walking. This is achieved by extending the prosthesis with an orthosis part and allows for part of the uncomfortable distal residual limb loading to be supported on the upperleg during ambulation. Furthermore, this allows for the amputee to have more control over -and sensory feedback from the prosthetic foot.

This chapter discussed the preliminary testing of two prototypes: The first prototypes' working was tested to get more insight in the mechanisms' working prior to clinical trials, using a Homo Sapiens Walking model and testing was done; the main question being whether or not plantarflexion could be established during the end of stance. The model consisted of a Homo Sapiens that was provided with braces underneath which prosthetic feet can be mounted (stilts). Subsequently the HSW-model walked on a treadmill and relative rotations of different segments in the foot were measured while the trial was video taped. Another prototype –which was basically a lighter version of the first- was designed to be tested on two TT-amputees in a laboratory environment while biomechanical data and feedback were gathered. Because in that stage the mechanisms behavior under walking condition loading was of great interest, both prototypes were used without shoes during measurements.

Preliminary measurements with HSW-models verified the hypothesized working of the 'pro-prototype' and nothing indicated that the proposed mechanisms' working needed to be adjusted before further developing it into a prototype suitable for clinical trials with TT-amputees (Chapter 3). The use of a HSW-model proved to be a quick and 'easy to use' tool for preliminary testing prototype foot characteristics while getting a feel of the prototypes' working during walking.

The preliminary testing with the two TT-amputees also show a correct –as hypothesized-working of the prosthetic foot: Plantarflexion -up to $\sim 18^\circ$ - under loading was measured while the affected ankle joint showed a significantly increased power-burst during the end of stance and biomechanical analysis of the gait patterns shows that energy is even generated about the affected ankle joint. Figure 4-19 shows the different phases of the prototypes working during a clinical session.

The effect of the prototypes' characteristics on the subjects gait during the final sessions resulted in an increased affected stance time for both subjects. Regarding this gait parameter, both subjects displayed improved symmetry during the final sessions. Walking velocity and sound step-length increased slightly for mr.D where mr.H displayed a slight decrease. Affected double support time increased for mr.H while sound double support time remained approximately the same as walking with his current prosthesis therefore improving symmetry. Mr.D however displayed a decrease in affected double support time while sound stance time increased. Although affected double support time increased while sound double support time

decreased during the final sessions, double support times remained more asymmetric than when mr.D was walking with his current prosthesis.

(a)

(b)

(c)

Figure 4-19: Shows the prototype working during ambulation. a) From Heel Strike until approximately Foot Flat. b) From Foot Flat until approximately Heel Off. c) From Heel Off until approximately Toe Off.

Both subjects show a decrease in the NME generated about the sound hip joint. The net mechanical energy absorbed about the sound hip decreased slightly for mr.H but increased slightly for mr.D. This can be a result of mr.D trying to control the increased plantarflexion action of the prosthetic foot during affected heel strike from the hip, increasing late stance sound hip loading (70% increase in peak late stance sound hip flexion torque and increased sound double support time). Both subjects also display a decrease in NME generated and absorbed about the sound knee, which concurs with the decrease in peak sound knee extension torques. Mr.H also displayed a decreased sound knee flexion torque during stance where it increased to average normal values for mr.D. The average NMEs generated about the sound ankle remain approximately the same or decrease where the NMEs absorbed show a slight increase. The overall decrease in NMEs generated and absorbed about the sound joints in combination with the overall decrease in loading, suggests an overall decrease in sound side loading, especially during sound weight acceptance.

Regarding the NMEs generated about the affected hip, mr.H displays an increase where mr.D displayed a decrease. Peak affected hip extension torques remained the same (mr.H) or decreased (mr.D). The NMEs absorbed about the affected hip slightly increased for mr.H and decreased for mr.D. The NME generated and absorbed about the affected knee decreased for mr.D. Mr.H showed a slight increase in affected knee NME absorption, which concurs with the increased peak affected knee extension torque during early stance in combination with an increased affected knee flexion rotation. Regarding the affected ankle joint, both subjects display a significant increase in NME generation (concurs with the increased power-burst during late stance). Mr.H also shows a significant increase in NME absorption where mr.D displays a decrease. Overall, the data suggest an increase in loading on the affected side, especially the affected knee.

In walking with the prototype both amputees felt a better push off and less resistance during the end of the affected stance compared to their current – leaf spring type- prosthesis. However, they were not able to determine whether or not this effect was an improvement for their walking due to the relative short wearing time compared to their current prosthesis.

Overall, the biomechanical data and patient feedback confirmed the hypothesized working of the proposed prototype prosthetic foot in which approximate center of gravity balancing seems a good guideline for determining the initial ankle joint stiffness.

When walking with the prototype foot in combination with the knee-ankle coupling during the final sessions, biomechanical data reveals little difference in joint kinetics and kinematics compared to walking with only the prototype foot. However, in walking with the knee-ankle coupling mr.H's affected knee extension torque increased where mr.D's affected knee torque altered to mainly flexor in direction. During the final sessions however both amputees felt that walking became less strenuous and more comfortable, showing the use of functional re-

establishing the knee ankle coupling working. The fact that an orthotic extension is required for the knee-ankle coupling remains an issue -from a cosmetic and easy donning and doffing point of view- that needs to be tackled before the knee ankle coupling will be worn outside a laboratory environment.

4.4 Appendix

4.4.1 Preliminary testing with HSW-models

Relevant measures HSW-models and test protocol

	Subject 1	Subject 2
Gender	male	male
Length	1.8 m	1.8 m
Weight	82 kg	94 kg
Shoe size	42	43

1. Put on harness (Fig.4-2a).
2. Walking during ± 1 min., subsequently adjusting of foot.
3. Static measurement (*HSW* model standing upright).
4. Begin measurement session: After walking comfortably at a certain speed on the treadmill the toe and ankle angles are measured during 30 sec.

Pro-prototype dimensions

Prototype dimensions (mm, °)	
<i>Foot</i>	
r_1	adjustable
r_2	adjustable
r_3	140
r_4	20
r_{h1}	45
r_{h2}	50
r_h	90
r_t	70
Angle $r_{h1} r_{h2}$	110
Angle $r_{h2} r_2$	150
C_1	Section 3.4.1
C_2	Section 3.2.1
<i>Knee-ankle coupling</i>	
r_a	60
r_k	30
Sh	600
C_3	Section 3.3.2

(b)

Figure 4-20: a) Dimensions of prototype used during the preliminary testing (Fig.4-2 and 4-3).
 (b) Table with the different dimensions. The two angles are for the unloaded prototype.

4.4.2 Relevant data of TT-amputees

<p>Mr. H</p> <p>Date of birth: 23-12-1934</p> <p>Right sided transtibial amputee because of trauma.</p> <p>Conventional socket (Support on upperleg with a cuff).</p> <p>Orthopedic shoe on sound side</p> <p>Active walker.</p> <p>Current prosthesis: Single axis foot (Otto Bock 1H38).</p> <p>Shoe size: 44</p>	
Mass (kg.)	82
Length (m)	1.85
R (m)	~1
Y (mm)	80-85
r_t (mm)	75-80
r_h (mm)	~85
r_3 (mm)	~145

Table 1: Relevant data on mr.H (see Figs.3-8; 3-9; and 3-25).

<p>Mr. D</p> <p>Date of birth: 24-12-1930</p> <p>Left sided transtibial amputee because of trauma.</p> <p>PTB suction socket.</p> <p>Active walker.</p> <p>Current prosthesis: Single axis foot (Otto Bock 1H32).</p> <p>Shoe size: 44</p>	
Mass (kg.)	100
Length (m)	1.87
R (m)	~1
Y (mm)	~80
r_t (mm)	~80
r_h (mm)	~90
r_3 (mm)	~145

Table 2: Relevant data on mr.D (see Figs.3-8; 3-9; and 3-25).

4.4.3 Biomechanical knee-ankle coupling data

Mr.H Setting A and knee ankle coupling

Figure 4-21: Joint kinematics and kinetics displayed during walking with the prototype in setting A in combination with the knee ankle coupling. RHIP, RKNE, and RANK are the Right hip, Right knee, and Right ankle joints respectively. LHIP, LKNE, and LANK are the Left hip, Left knee and Left ankle joints respectively. RFy and RFx are the vertical and horizontal (in the plane of progression) Ground Reaction Force components established by the right leg respectively. LFy and LFx are the vertical and horizontal (in the plane of progression) Ground Reaction Force components established by the left leg respectively.

Mr.H Setting B and knee ankle coupling

Figure 4-22: Joint kinematics and kinetics displayed during walking with the prototype in setting B in combination with the knee ankle coupling. RHIP, RKNE, and RANK are the Right hip, Right knee, and Right ankle joints respectively. LHIP, LKNE, and LANK are the Left hip, Left knee and Left ankle joints respectively. RFy and RFx are the vertical and horizontal (in the plane of progression) Ground Reaction Force components established by the right leg respectively. LFy and LFx are the vertical and horizontal (in the plane of progression) Ground Reaction Force components established by the left leg respectively.

Mr.D setting A and knee ankle coupling

Figure 4-23: Joint kinematics and kinetics displayed during walking with the prototype in setting A in combination with the knee ankle coupling. RHIP, RKNE, and RANK are the Right hip, Right knee, and Right ankle joints respectively. LHIP, LKNE, and LANK are the Left hip, Left knee and Left ankle joints respectively. RFy and RFx are the vertical and horizontal (in the plane of progression) Ground Reaction Force components established by the right leg respectively. LFy and LFx are the vertical and horizontal (in the plane of progression) Ground Reaction Force components established by the left leg respectively.

Mr.D setting B and knee ankle coupling

Figure 4-24: Joint kinematics and kinetics displayed during walking with the prototype in setting B in combination with the knee ankle coupling. RHIP, RKNE, and RANK are the Right hip, Right knee, and Right ankle joints respectively. LHIP, LKNE, and LANK are the Left hip, Left knee and Left ankle joints respectively. RFy and RFx are the vertical and horizontal (in the plane of progression) Ground Reaction Force components established by the right leg respectively. LFy and LFx are the vertical and horizontal (in the plane of progression) Ground Reaction Force components established by the left leg respectively.

5 Concluding Remarks

Gait studies reveal that adding a more natural –anatomic- ankle functionality to conventional prosthetic feet is beneficiary for TT-amputee walking: This includes a compliant plantarflexion ankle rotation during affected heel strike and allowing for a controlled dorsalflexion of the lowerleg relative to the prosthetic foot during ambulation. Both added features seem to enable a more smooth progression of the trunk during walking, improving the amputees' perception of walking comfort, control and cosmetics. However, these studies also suggest excessive straining of remaining structures due to the lack of affected plantarflexion action during the end of affected stance, leveling down the 3C-level and increasing secondary health risks in the long run. Therefore adding a plantarflexion action to current prosthetic feet seems beneficiary for TT-amputee walking. The plantarflexion action acts twofold: The plantarflexion of the foot increases affected leg-length and secondly generates enough energy to enable a more smooth progression of the oscillating body segments as energy is being transferred from affected to the sound side, via the trunk.

Chapters 1 to 4 of this thesis were about the development –and preliminary testing of such a plantarflexing prosthetic foot. The foot was developed using an anthropomorphic approach and typical sagittal anatomic ankle-foot characteristics were used as a guideline for determining the initial prosthetic ankle-foot characteristics.

A plantarflexing foot is proposed, consisting of a four bar linkage in combination with a flexible element. The flexible element stiffness is determined such that gravity acting on the center of gravity of the amputee, rotating at a distance R about the ankle joint, is approximately statically balanced. It has been shown numerically that such a mechanism approaches typical anatomic ankle-foot characteristics under walking conditions, but preliminary clinical testing has also been done with two TT-amputees in a laboratory environment in which the amputees walked without a shoe on the affected side. The preliminary testing results reveal the hypothesized working of the foot.

Both amputees felt an increased push-off and less resistance during the end of stance, the short wearing time compared to their current prosthesis prohibited them to conclude whether the new prosthetic characteristics were an improvement on their current prosthesis or not; this clearly requires the prototype to be worn outside a laboratory environment (with shoe on affected side) for longer periods of time prior to clinical trials.

In the proposed prosthetic foot, the plantarflexing functionality normally present in the anatomic foot during the end of stance is restored. Since the preliminary clinical testing results are promising, further research can focus on the influence of frontal and transversal plane movements in the ankle-foot prosthesis. Also, the first steps towards incorporating the

HEELfoot functionality in an ankle-foot orthosis have already been taken[†]. Such an orthosis can be beneficiary for persons with –temporary- plantarflexor dysfunction (e.g. due to stroke or injury).

Beside the plantarflexing prosthetic foot, a knee-ankle coupling is re-established during walking to further improve on walking comfort. The aim is to reduce uncomfortable distal residual limb loading while increasing control over the prosthetic foot from the knee and establishing a more natural energy/ force transfer between anatomic knee and prosthetic ankle. Again typical normal walking characteristics were used as a guideline for determining initial coupling setting.

In walking with the knee-ankle coupling it is seen that each amputee reacts differently on the knee-ankle coupling regarding the affected knee; one displays increased knee-extensor torques where the other displays increased affected knee-flexor torques. However, both remark that walking becomes less strenuous and laborious when walking with the knee-ankle coupling. Besides these immediate (hypothesized) effects there are more advantages: The orthotic extension –with the right design- also increases frontal plane affected knee stability during weight acceptance while decreasing residual limb loading (decreasing the chance on long-term health risks).

The potential advantages of the knee-ankle coupling for a TT-amputee are evident. The main challenge with the knee-ankle coupling at this point in time lies in achieving a cosmetically sound (easy donning and doffing) design so that it can be worn on a daily basis outside the laboratory environment.

[†]Janssen E.C. (2002) A plantarflexing ankle-foot orthosis, Masters thesis, Report BW-140, Twente University, Enschede

SECTION 2:

Static gravity balancing for rotations about a fixed pivot

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Static gravity balancing for rotations about a fixed pivot

Originally the project was funded for two researchers: One who would dedicate time to describe and find applications for static balancing mechanisms in the biomechanical engineering field and one who was more dedicated to the design of a new prosthetic foot. This section discusses several balancing principles focussed on the conception of a static gravity balancing ankle joint. The Center Of Gravity of a subject standing in an upright position –and on one leg- is then approached as an inverted pendulum rotating at a distance R about the affected ankle joint.

1.1 Introduction

In mechanical engineering three kinds of energy are known: First there is the potential energy, which can be divided into elastic and gravitational energy. Secondly, when an object with a certain mass is moving, it is said to possess kinetic energy. Finally, with all movements some energy is lost (heat/ noise) caused by hysteresis. This last form of energy is often neglected and considered unwanted (but actually inevitable in most constructions). When a system is considered statically, the kinetic energy of the mechanism is not considered. During the course of this chapter all mechanisms will be considered statically with hysteresis neglected.

Statically considered, a position of a mechanism is in stable equilibrium if the potential energy of the mechanism rises for all movements away from that position. If the potential energy of the mechanism drops for movements away from that position, the mechanism is unstable. When the potential energy remains constant, the mechanism is said to be statically balanced (in a constant state of equilibrium). To determine the state of equilibrium (unstable, stable or statically balanced) of a certain position p , generally the potential energy of a mechanism as a function of the movement $V(m)$ is developed into a Taylor progression around that position³:

$$V(m)|_p = V(p) + \frac{\partial V(p)}{\partial m} \frac{(m-p)}{1!} + \frac{\partial^2 V(p)}{\partial m^2} \frac{(m-p)^2}{2!} + \dots + \frac{\partial^n V(p)}{\partial m^n} \frac{(m-p)^n}{n!}$$

If stable or statically balanced equilibrium is demanded (potential energy rise or constant) it is easily seen that:

$$\frac{\partial V(p)}{\partial m} \frac{(m-p)}{1!} + \frac{\partial^2 V(p)}{\partial m^2} \frac{(m-p)^2}{2!} + \dots + \frac{\partial^n V(p)}{\partial m^n} \frac{(m-p)^n}{n!} \geq 0$$

The mechanism can only be in equilibrium if the first derivative of the potential energy at position p is equal to zero. After all, if the first derivative is not equal to zero a movement m

from p can always be found that causes the potential energy of the mechanism to drop. If the second derivative now is positive, this is a stable equilibrium, if negative an unstable equilibrium. With the second derivative zero, the third also has to be zero (same reasoning as for the first derivative). Subsequently the fourth has to be positive to cause a potential energy rise for every movement m away from p (stable equilibrium). This process is continued until an (un)stable equilibrium for a position p is found. Note however that when a mechanism is statically balanced, every derivative is zero because the potential energy characteristic is a constant. Using the method set out above, it is easily seen that the pendulum in Fig.1-1a is in stable and the pendulum in Fig.1-1b is in unstable equilibrium:

$$V_{m_1}(\theta) = -m_1 g R_1 \cos \theta \Rightarrow \left. \frac{\partial V(\theta)}{\partial \theta} \right|_{\theta=0} = 0 \Rightarrow \left. \frac{\partial^2 V(\theta)}{\partial^2 \theta} \right|_{\theta=0} > 0$$

$$V_{m_2}(\theta) = m_2 g R_2 \cos \theta \Rightarrow \left. \frac{\partial V(\theta)}{\partial \theta} \right|_{\theta=0} = 0 \Rightarrow \left. \frac{\partial^2 V(\theta)}{\partial^2 \theta} \right|_{\theta=0} < 0$$

Figure 1-1: A pendulum: a) In stable equilibrium. b) In unstable equilibrium. c) Statically balanced with counterweight. d) Statically balanced with spring.

Gravity acting on the pendulums' mass can be statically balanced through the use of a counterweight or a spring. Statically seen, this means that the mass can be rotated without any work. However dynamically seen, work still has to be done to de –or accelerate the mass (although these energies can also compensate each other) according to:

$$M = I \cdot \ddot{\theta}$$

With M the torque needed to achieve angular acceleration $\ddot{\theta}$ of a body with a mass moment of inertia I .

Although depending on type of spring and the position of spring attachment points in some cases, generally it can be stated that employing a spring to statically balance a rotating mass will result in a lighter construction than when a counterweight is employed. Dynamically seen

this can have the extra advantage of a decrease in mass moment of inertia, causing the amount of work to de- or accelerate the mass to decrease (allows for lightweight actuators). Regardless of inertia however, lightweight constructing in biomechanical engineering is generally required (especially in prosthetic or orthotic designing). The application of springs to balance parasitic spring forces or masses in a mechanism instead of other masses can –if employed properly– therefore be beneficiary for biomechanical designs.

In the biomechanical engineering field, spring-to-spring balancing principles are commonly applied to compensate for unwanted (parasitic) elastic forces in mechanisms. A nice example of such an application is the use of a compensation spring in a prosthetic hand to approximately balance the unwanted forces caused by the compliant properties of cosmetic glove coverings (Fig.1-2a)⁷. Another example is shown in Fig.1-2b. The displayed mechanism is a pair of forceps with compliant jaws. A spring is added to the handle of the forceps to statically balance the forces caused by the compliant jaws so that the operating force is near zero. A mechanism of Rolling Links is used to minimize hysteresis in the force transmission chain. A pair of such forceps improves the sensory feedback of gripping forces occurring at the jaws during minimal invasive surgery.⁷

Figure 1-2: Examples of static spring-to-spring balancing mechanisms in the biomechanical engineering field (copied from Herder⁷): a) Prosthetic hand with an extra spring to balance the forces caused by the compliance of the cosmetic glove covers (represented by a dashed spring). b) Forceps that enable a surgeon to have a maximal sensation of the forces the gripper applies during surgery. The instrument is equipped with an extra spring to compensate for the forces caused by the compliance of the gripper.

Gravity-balancers in the biomechanical engineering field are primarily used to compensate for the weight of weakened or paralyzed limbs. An example of a highly appreciated static gravity balancing mechanism is the elbow orthosis shown in Fig.1-3. The orthosis is intended to approximately balance the weight of the forearm with a spring. A slight underbalance causes the forearm to hang down when the upper-arm is alongside the trunk. However, when the upper arm is abducted, the spring action overcomes gravity, causing the forearm to lift up. A

ratchet mechanism locks the forearm at 90° and unlocks at repeated action. This makes it possible for the patient to control elbow flexion with the shoulder muscles.²

Figure 1-3: Wilmer elbow orthosis²: (a) photograph of a patient wearing the orthosis, (b) photograph of the orthosis, (c) diagram showing the rubber spring which counteracts the weight of the forearm.

Figure 1-4: Schematic representation of prosthetic ankle-foot combination and the movement of the Center Of Gravity. The torsion spring in the ankle joint has a rotational stiffness such that gravity acting on the COG -at a distance R from the ankle joint- is statically balanced (for ankle dorsalflexion).

Section 1 of this thesis presented a prosthetic foot concept consisting of a four-bar-linkage in combination with a Center Of Gravity Balancing ankle joint (for dorsalflexing rotations of the lowerleg relative to the foot). Such a mechanism can approximate the typical anatomic ankle characteristics occurring during walking at Self-selected Walking Speed. In that case, gravity

balancing is used as a guideline for determining initial prosthetic ankle joint rotational stiffness for a prosthetic walker (Fig.1-4).

This section focuses on the static balancing of gravity acting on an inverted pendulum by means of springs. The aim was to study whether or not mechanically simple perfect gravity balancing mechanisms exist or can be conceived, suitable for further development into a prosthetic ankle joint. In this study both 1 degree of freedom (dof) as well as 2dof balancing was considered. Using the potential energy consideration set out earlier, initially balancing with ideal springs (free-length $l_0=0$) is considered; subsequently balancing with technical springs (free-length $l_0 \neq 0$) is discussed.

1.2 Static gravity balancing with ideal springs

Figure 1-5: The force-deflection curves of three linear springs with equal stiffness C . The springs are stretched between predetermined points A and B : a) An ideal spring. b) A technical spring with zero length l_0 . c) A linear technical spring used as an ideal spring by positioning the zero length l_0 outside the attachment points A and B . d) By using special coiling techniques a technical spring can be pre-stressed such that it can be used as an ideal spring between the predefined attachment points for a limited range.

Prior to discussing spring to spring –or gravity balancing mechanisms it is important to explain what is meant with so called ‘ideal springs’ and ‘technical springs’ since all static balancing mechanisms are conceived with one of these kind of springs. Ideal springs have a zero free-length, which means that the unloaded length of these springs is equal to zero ($l_0=0$). To clarify what is meant, Fig.1-5a shows the characteristic of a linear ideal spring where Fig.1-5b shows the characteristic of a linear technical spring ($l_0 \neq 0$) when stretched between two predefined

attachment points A and B. It is of course possible to use technical springs as ideal springs by placing the free-length of the technical spring outside the predefined attachment points (Fig.1-5c) or by using special coiling techniques to pre-strain the technical springs (Fig.1-5d). Note that a pre-strained technical spring only approximates the ideal spring in for a limited range (l_0 to B). For more examples on technical springs used as ideal springs the reader is referred to the literature.^{6,7, 8, 10, 12}

This section will show that perfect static gravity balance is possible using the basic set-up of an initially static spring-to-spring balancer. First the basic architecture of a 1dof static spring-to-spring balancing mechanism employing ideal springs will be discussed. Subsequently, using a potential energy consideration, the geometrical conditions for perfect static balance are derived, followed by a consideration of what happens if these conditions are not satisfied. Subsequently, using the same approach, similar conditions will be derived for a 2dof (allows for 2 axis of rotation) mechanism employing three ideal springs.

1.2.1 1 dof static gravity balancing

The basic 1dof gravity balancing mechanism shown in Fig.1-6 consists of a mass, rotating at a distance R about a fixed pivot and an ideal linear spring with a stiffness C connected to the ground and to rod R . The mechanism is exactly statically balanced when the condition given in eq.2) is satisfied.^{4, 6, 8, 10, 12, 16}

Figure 1-6: a) A mass rotating about a fixed pivot is perfectly statically balanced by means of an ideal spring. b/ c) Static balance with technical springs.

The total potential energy expression of the mechanism is the sum of the potential energy of the mass V_m and the spring V_s :

$$V_{tot} = V_s + V_m = \frac{1}{2} C(h^2 + i^2) - Chi \cos \gamma + mgR \cos \gamma \quad 1)$$

It is readily seen that the potential energy remains constant when:

$$Chi = mgR \quad 2)$$

Figure 1-7: Basic static spring-to-spring balancer configuration. The mechanism consists of two triangles with apical angles of 2α and 2β , interconnected with a hinge joint at the apexes. The mechanism is pre-strained with two ideal springs ($l_0=0$) with stiffness C_1 and C_2 and a rotation of 2θ is enforced on the top triangle.

Another –more general– method of deriving a static gravity balancer with ideal springs is adjusting the geometry of an initially static spring-to-spring balancer. The mechanism shown in Fig.1-7 can be seen as the basic set-up for the 1dof static spring-to-spring balancer amply discussed in literature.^{7, 9, 10} The mechanism consists of two triangles with apical angles of 2α and 2β . The triangles are interconnected with a hinge joint at the apexes. Two ideal linear springs ($l_0=0$) with stiffness C_1 and C_2 are connected to the ends of lever-arms a/b and c/d respectively, and the top triangle is rotated through an angle 2θ . The potential energy of the mechanism is:

$$V = V_1 + V_2 = \frac{1}{2} C_1 \ell_1^2 + \frac{1}{2} C_2 \ell_2^2 \quad 3)$$

With ℓ_1 and ℓ_2 the spring lengths and C_1 and C_2 the spring stiffness.

Using the cosine rule, and rearranging gives:

$$V = \frac{1}{2} C_1 (a^2 + b^2) - C_1 ab \cos(\psi_1) + \frac{1}{2} C_2 (c^2 + d^2) - C_2 cd \cos(\psi_2) \quad 4)$$

With $\psi_1 = \pi - \alpha - \beta + 2\theta$; $\psi_2 = \pi - \alpha - \beta - 2\theta$ and $\cos(\alpha + \beta) = \cos \alpha \cos \beta - \sin \alpha \sin \beta$

Equation 4 can be written as:

$$\begin{aligned}
 V = & \frac{1}{2} C_1 (a^2 + b^2) + C_1 ab (\cos 2\theta \cos(\alpha + \beta) + \sin 2\theta \sin(\alpha + \beta)) \\
 & + \frac{1}{2} C_2 (c^2 + d^2) + C_2 cd (\cos 2\theta \cos(\alpha + \beta) - \sin 2\theta \sin(\alpha + \beta))
 \end{aligned} \tag{5}$$

Now it is easily seen that the potential energy dependency on θ vanishes from the eq.5) if the conditions set out in eq.6) are satisfied:

$$C_1 ab = C_2 cd \text{ and } \alpha + \beta = \frac{1}{2}\pi \tag{6}$$

Using the conditions set out in eq.6) an infinite number of statically balancing spring-to-spring mechanisms (combinations) can be derived.

Figure 1-8: Mechanism as Fig.1-7 with $a=b=c=d$ and $\alpha=\beta=0.25\pi$. The figure shows two Rolling Link versions of the basic 1dof two-spring mechanism. The apexes of the triangles can be shifted apart (a) or overlapped (b) with a distance l_{sh} without changing the mechanisms characteristics, provided springs with a free-length of respectively $l_0=l_{sh}$ or $l_0=-l_{sh}$ are employed. Note however, that a full 360° rotation will not be possible anymore.

It is noteworthy that in the special case when $a=b \neq c=d$, $\alpha=\beta$ and the conditions set out in eq.6) satisfied, the apexes of the triangles of the described mechanism can be shifted apart^{1, 7} or overlapped¹³ without changing the characteristics of the mechanism. However, to ensure the same working of the spring-triangle mechanism, springs with a free-length l_0 have to be employed to compensate for shifting the triangle apexes with a distance l_{sh} such that $l_0=l_{sh}$ for shifting the apexes apart with, and $l_0=-l_{sh}$ for overlapping the triangles with l_{sh} (Fig.1-8).

Figure 1-8 shows a mechanism with $a=b=c=d$ and $C_1=C_2=C$. The figure shows that overlapping/ shifting apart the triangles allows for the easy application of so called Rolling Link Mechanisms to ensure little friction.¹¹ The advantage of the first mechanism is that actual technical springs with free-length $l_0=d$ can be employed. The second mechanism also is self-adjusting (in the sense that it finds its own static balancing position) and compact. A disadvantage is that that these mechanisms do not allow for full 360° rotations.

Figure 1-9: Two similar unstable mechanisms: a) A linear ideal spring stretched between two lever arms h and i ; and h and j . b) Two masses rotating about a fixed pivot.

If the conditions for perfect static spring-to-spring balance are not satisfied, both a $\cos 2\theta$ and a $\sin 2\theta$ term remain in the potential energy expression set out in eq.5) the mechanism is no longer statically balanced. To achieve a static balancing mechanism again, another mechanism has to be mounted to the spring mechanism which also has a potential energy expression built up out of constants times a $\sin 2\theta$ and a $\cos 2\theta$ term. Only then the θ dependency can be eliminated again from the *total* potential energy expression.

With that in mind, the potential energy characteristic of the two masses rotating through an angle 2θ about a fixed pivot as shown in Fig.1-9b, can be described as:

$$V_m = m_1 g R_1 \cos 2\theta - m_2 g R_2 \sin 2\theta \quad 7)$$

Whereas the potential energy characteristic of the two ideal springs, stretched over the rods h , i , and j , rotating through an angle 2θ about a fixed pivot as shown in Fig.1-9a, is described as:

$$V_s = \frac{1}{2} C_3 (h^2 + i^2) + C_3 h i \cos 2\theta + \frac{1}{2} C_4 (h^2 + j^2) - C_4 h j \sin 2\theta \quad 8)$$

It is seen that both eq.7) and eq.8) are built up out of constant, $\cos 2\theta$ and $\sin 2\theta$ terms. Therefore perfect static balance of the basic two-spring mechanism can be re-established through mounting two extra ideal springs or masses to the top triangle as shown in Fig.1-10a and 1-10b respectively, if the conditions set out in eq.6) are not satisfied. The conditions for perfect static balance then become:

*spring-to-spring balancing*9)

$$C_1ab = C_2cd \text{ and } \alpha + \beta = 0.5\pi$$

additional ideal springs

$$C_3hi = -(C_1ab + C_2cd)\cos(\alpha + \beta) \text{ and } C_4hj = (C_1ab - C_2cd)\sin(\alpha + \beta)$$

additional masses

$$m_1gR_1 = -(C_1ab + C_2cd)\cos(\alpha + \beta) \text{ and } m_2gR_2 = (C_1ab - C_2cd)\sin(\alpha + \beta)$$

Figure 1-10: Two perfect static balancing mechanisms. The basic spring-to-spring mechanism is as shown in Fig1-7 but with apical angles $\alpha=\beta\neq 0.25\pi$. To re-establish perfect balance, extra ideal springs and/or masses are mounted to the top triangle of the spring mechanism.

Using the approach set out above also allows for several combinations of masses in combination with –extra– springs to be mounted on the two-spring mechanism shown in Fig.1-7 (instead of only two extra masses or two extra springs). It is noteworthy however that after adding ‘extra masses’ -or ‘extra springs’ to the basic two-spring mechanism (to re-establish perfect balance), the resulting mechanism does not allow for shifting apart or overlapping the apexes of the triangles with a distance l_{sh} (through employing technical springs with a free-length $l_0=l_{sh}$) as was the case for the two-spring balancing mechanism (Fig.1-7). After all, the

masses –or extra springs then no longer rotate about one pivot but about two pivots with two links such that the rotation of θ of link R_2 ($=l_{sh}$) causes a rotation of 2θ of link R_1 (Fig.1-11a and b). Thus the conditions set out in eq.9) are no longer valid. In advance: Although not useful now, this rotational link relation will prove to be useful when perfect gravity balancing with technical springs is aspired (section 1.3.1).

Figure 1-11: A gravity balancing mechanism satisfying eq.9), with the basic spring-to-spring mechanism according to Fig.1-7 with $a=b=c=d$ and $\alpha=\beta\neq 0.25\pi$. By shifting the apexes of the triangles apart the mass rotates about two pivots with two links R_1 and R_2 such that a rotation of θ of R_2 causes a rotation of 2θ of R_1 therefore the conditions set out in eq.9) are no longer valid.

1.2.2 2 dof static balancing

The figure below shows a mass rotating about a pivot where the hinge is replaced with a spherical joint; allowing for 3dof; φ , θ and ψ . In the course of this section a spring mechanism will be derived that statically balances the mass for rotations about an arbitrary chosen axis lying in the axis the xy -plane (balanced for 2dof; θ and φ). This will be done using a similar method as set out in the previous section. First the geometric balancing conditions of a 2dof – ideal- spring-to-spring balancer will be derived, followed by a consideration of the consequences if these conditions are not satisfied.

Consider the spring-mechanism shown in Fig.1-12: The mechanism consists of two cones with apical angles of 2α and 2β . The cones are interconnected with a spherical joint -in stead of cross coupling like joint- in the apex. On the surfaces of the cones, six lever arms (*a to f*) are mounted such that in the neutral position, the projection lines of the upper -and lower cone levers on the xy -plane coincide with mutual angles of 120° . Finally, three *ideal* linear springs with stiffness C_1, C_2, C_3 are spanned between the ends of respectively the lever arms *a* and *b*; *c* and *d*; *e* and *f*. The bottom cone and the coordinate system are considered fixed to the ground. During the course of this section the conditions for perfect static balance for rotations of the top cone about an arbitrary chosen axis in the xy -plane will be derived. Subsequently the torsion stability of these static-balancing positions is considered for possible -subsequent- rotations about the top cones' axis of revolution. Since this is done by means of potential energy considerations, the way the mechanisms' potential energy is considered during the following sections will first be explained.

Every position of the top cone relative to the bottom cone can be described by a rotation of the top cone about an arbitrary chosen axis in the xy -plane (through the spherical joint) followed by an axial rotation about its' axis of revolution (Fig.1-12c and d respectively). This allows for the following mechanism consideration: Instead of rotating the top cone about an arbitrary axis in the xy -plane, the mechanism -as a whole- is rotated through an (arbitrary chosen) angle ϕ about the z -axis (Fig.1-12a). Subsequently the bottom cone is again considered fixed to the ground and a rotation of 2θ is enforced on the top cone about the x -axis (Fig.1-12b→Fig.1-12c). Finally a rotation ψ is enforced on the top cone about its axis of revolution while maintaining the rotation about the x -axis (Fig.1-12c→Fig.1-12d).

Figure 1-12: a) 3D -and top view of mechanism consisting of 6 levers, three ideal springs and two cones interconnected with a spherical joint in the apexes. The mechanism is in its neutral position. b) The cone in neutral position. c) The top cone rotated through an angle 2θ about the x-axis, relative to the bottom cone. d) As c) but with the top cone axially rotated about its normal.

Using vector notation, the paths of the spring attachment points are described by the lever-arm vectors relative to a global coordinate system, which is also considered fixed to the ground (Fig.1-13). With the mechanism in the neutral position and rotated –as a whole- through an angle φ about the z -axis, it is seen that the lever-arm vectors relative to the xyz -coordinate system are given by (Fig.1-13 and Fig.1-12):

$$\begin{aligned}\bar{v}_a &= \begin{bmatrix} x \\ y \\ z \end{bmatrix} = \begin{bmatrix} -a \cdot \sin \alpha \cdot \sin \varphi \\ -a \cdot \sin \alpha \cdot \cos \varphi \\ a \cdot \cos \alpha \end{bmatrix}; \quad \bar{v}_b = \begin{bmatrix} -b \cdot \sin \beta \cdot \sin \varphi \\ -b \cdot \sin \beta \cdot \cos \varphi \\ -b \cdot \cos \beta \end{bmatrix}; \quad \bar{v}_c = \begin{bmatrix} -c \cdot \sin \alpha \cdot \sin(\frac{1}{3}\pi - \varphi) \\ c \cdot \sin \alpha \cdot \cos(\frac{1}{3}\pi - \varphi) \\ c \cdot \cos \alpha \end{bmatrix}; \\ \bar{v}_d &= \begin{bmatrix} -d \cdot \sin \beta \cdot \sin(\frac{1}{3}\pi - \varphi) \\ d \cdot \sin \beta \cdot \cos(\frac{1}{3}\pi - \varphi) \\ -d \cdot \cos \beta \end{bmatrix}; \quad \bar{v}_e = \begin{bmatrix} e \cdot \sin \alpha \cdot \sin(\frac{1}{3}\pi + \varphi) \\ e \cdot \sin \alpha \cdot \cos(\frac{1}{3}\pi + \varphi) \\ e \cdot \cos \alpha \end{bmatrix}; \quad \bar{v}_f = \begin{bmatrix} f \cdot \sin \beta \cdot \sin(\frac{1}{3}\pi + \varphi) \\ f \cdot \sin \beta \cdot \cos(\frac{1}{3}\pi + \varphi) \\ -f \cdot \cos \beta \end{bmatrix}\end{aligned}\quad (10)$$

The position of the vectors on the top cone relative to the xyz -coordinate system after a subsequent rotation 2θ about the x -axis followed by a rotation about the local z -axis (z') is described by the following matrix operation (Fig.1-12):

$$\bar{v}' = \begin{bmatrix} x' \\ y' \\ z' \end{bmatrix} = [R_x] \bullet [R_{z'}] \bullet \bar{v} = [R_x] \bullet [R_{z'}] \bullet \begin{bmatrix} x \\ y \\ z \end{bmatrix} \quad (11)$$

With: $[R_{z'}]$ and $[R_x]$:

$$[R_{z'}] = \begin{bmatrix} \cos \psi & \sin \psi & 0 \\ -\sin \psi & \cos \psi & 0 \\ 0 & 0 & 1 \end{bmatrix} \quad \text{and} \quad [R_x] = \begin{bmatrix} 1 & 0 & 0 \\ 0 & \cos 2\theta & \sin 2\theta \\ 0 & -\sin 2\theta & \cos 2\theta \end{bmatrix} \quad (12)$$

With ψ or θ equal to zero, the rotation matrix $[R_{z'}]$ and $[R_x]$ respectively become the unit matrix.

The vectors that span two mutual spring attachment points are the difference of the bottom cone vector and the top cone vector. The length of this vector $\bar{v}_{length}$, is the distance between the two spring attachment points.

$$\begin{aligned}\bar{v}_b - \bar{v}_a' &= \bar{v}_{length_1} \\ \bar{v}_d - \bar{v}_c' &= \bar{v}_{length_2} \\ \bar{v}_f - \bar{v}_e' &= \bar{v}_{length_3}\end{aligned}\quad (13)$$

Because a potential energy consideration is used to determine the perfect static balance conditions of the mechanism, it is important to recognise that the potential energy of a spring is determined as a constant times the square of the spring elongation. The cones and lever-arms are considered to be rigid and without mass thus the total potential energy of the mechanism is the sum of the potential energies of the three springs. Since ideal springs are employed, the total potential energy of the mechanism is therefore given by:

$$V_{tot} = \frac{1}{2} C_1 \cdot \bar{v}_{length_1}^T \bullet \bar{v}_{length_1} + \frac{1}{2} C_2 \cdot \bar{v}_{length_2}^T \bullet \bar{v}_{length_2} + \frac{1}{2} C_3 \cdot \bar{v}_{length_3}^T \bullet \bar{v}_{length_3} \quad (14)$$

Figure 1-13: a) Vectors of lever arms of the 2dof mechanism, relative to the coordinate system in the spherical joint.

With the general potential energy expression determined, the static balancing conditions can be determined for rotations about an arbitrary chosen axis in the xy -plane and followed by a torsion stability consideration.

Rotation top cone about an axis in the xy -plane

In order for the mechanism to be perfectly statically balanced for a rotation of the top cone relative to the bottom cone about an axis lying in the xy -plane, the total potential energy must remain constant. Since ideal springs are used, the distance between two spring attachment points is equal to the strained spring-length and therefore the measure for the potential energy stored in the spring. The length of each spring relative to the coordinate system is determined according to (Fig.1-14):

$$l^2 = \Delta x^2 + \Delta y^2 + \Delta z^2 \quad \text{and} \quad V_s = \frac{1}{2} C \cdot l^2$$

However, when considering this mechanism, it is seen that rotations of the top cone about the x -axis only cause the Δy –and Δz -components to change, where the Δx -component remains constant in the spring length determination. Therefore it is possible to only consider the projection of the spring-lengths on the yz -plane for the “potential energy remains constant consideration”. Then it is not necessary to apply vector notation and multiplication and the spring-length variation can be directly derived from applying the different basic goniometric rules (similar to the 1dof mechanisms).

Figure 1-14: Determining of a spring length in the coordinate system.

With a subsequent rotation about the local z' -axis however, such a consideration is no longer valid since spring-length variation occurs in all directions. Then it is more convenient to use vector and/ or matrix notation and multiplication. To maintain some sort of uniformity in the ‘main consideration line’, the ‘cosine rule consideration’ is only recorded in the Appendix (section 1.6.1) and the ‘vector consideration’ is set out in the following section for both – subsequent- rotations.

Only considering rotations in the xy -plane, ψ is considered zero and $[R_z]$ becomes the unit matrix. Subsequently following the matrix-vector multiplying procedure set out above, the total potential energy of the spring mechanism can be determined as a function of φ and θ . After this matrix-vector multiplication the resulting equation is written out. Subsequently the terms $C_1 \cdot ab$, $C_2 \cdot cd$, $C_3 \cdot ef$, $\cos^2 \varphi$, $\sin \varphi \cos \varphi$ and $\cos \varphi$ are collected. This is all done symbolically with *Maple*: The resulting total potential energy V_{tot} expression of the mechanism becomes:

$$\begin{aligned}
 V_{tot} = & (C_1 \cdot ab - \frac{1}{2} C_2 \cdot cd - \frac{1}{2} C_3 \cdot ef) \cdot \cos^2 \varphi \cdot \sin \beta \cdot \sin \alpha \cdot (1 - \cos 2\theta) \\
 & + (C_2 \cdot cd - C_3 \cdot ef) \sin \beta \sin \alpha (1 - \cos 2\theta) \cdot \frac{1}{4} \sqrt{3} \sin 2\varphi \\
 & + (C_1 \cdot ab - \frac{1}{2} C_2 \cdot cd - \frac{1}{2} C_3 \cdot ef) \sin(\alpha + \beta) \cdot \cos \varphi \cdot \sin 2\theta \\
 & - (C_2 \cdot cd - C_3 \cdot ef) \sin(\alpha + \beta) \cdot \frac{1}{2} \sqrt{3} \sin \varphi \cdot \sin 2\theta \\
 & + 2 \cos \beta \cos \alpha \cos^2 \theta \cdot C_1 \cdot ab + \left(-\frac{3}{2} \sin \beta \sin \alpha + 2 \cos \beta \cos \alpha \right) \cos^2 \theta \cdot C_2 \cdot cd \\
 & + \left(-\frac{3}{2} \sin \beta \sin \alpha + 2 \cos \beta \cos \alpha \right) \cos^2 \theta \cdot C_3 \cdot ef + K
 \end{aligned} \tag{15}$$

With K the constant terms in the equation.

The φ -dependency can only be eliminated from the potential energy expression when $C_1 \cdot ab = C_2 \cdot cd = C_3 \cdot ef$. In other words: Then the mechanism is rotation symmetric. In doing so the potential energy also expression reduces to:

$$\begin{aligned}
 V_{tot} = & (-3 \sin \beta \sin \alpha + 6 \cos \beta \cos \alpha) \cos^2 \theta \cdot C_1 \cdot ab + K \Rightarrow \\
 V_{tot} = & -\frac{3}{2} \cos 2\theta \cos \alpha \cos \beta (\tan \alpha \tan \beta - 2) \cdot C_1 \cdot ab + K
 \end{aligned} \tag{16}$$

With K the constant terms in the equation.

Equation 16 reveals that the potential energy of the mechanism remains constant for every rotation 2θ of the top cone when the following conditions are satisfied:

$$C_1 \cdot ab = C_2 \cdot cd = C_3 \cdot ef \quad \text{and} \quad \tan \alpha \tan \beta = 2 : \tag{17}$$

As was also the case for the 1dof mechanism, it is seen that the product of a constant and a $\cos 2\theta$ term remains in the expression if $\tan \alpha \tan \beta \neq 2$. This $\cos 2\theta$ term can be zeroed out using a similar approach as for the 1dof mechanism (Figs.1-15). However, since only a $\cos 2\theta$ remains, the mass m_2 and the spring stiffness C_4 (responsible for the sine terms) are zero. Thus by mounting a mass to the extension of the axis of revolution of top cone or stretching an extra ideal spring over the extensions of the axis of revolutions of both the top –and the bottom

cone, the mechanism can be statically balanced again. Similar to the 1dof mechanism, two extra conditions for establishing perfect static balance can therefore be derived:

spring-to-spring balancing 18)

$$C_1 \cdot ab = C_2 \cdot cd = C_3 \cdot ef \quad \tan \alpha \tan \beta = 2$$

additional ideal spring

$$C_1 \cdot ab = C_2 \cdot cd = C_3 \cdot ef \quad \text{and} \quad C_s hi = -\frac{3}{2} C_1 ab \cos \alpha \cos \beta (2 - \tan \alpha \cdot \tan \beta)$$

additional mass

$$C_1 \cdot ab = C_2 \cdot cd = C_3 \cdot ef \quad \text{and} \quad mgR = -\frac{3}{2} C_1 ab \cos \alpha \cos \beta (2 - \tan \alpha \cdot \tan \beta)$$

Figure 1-15: Mechanism as shown in Fig.1-13 but with apical angles such that $\tan \alpha \tan \beta \neq 2$ with: a) extra ideal spring with stiffness C_s . b) extra mass m to re-establish perfect static balance (eq.18).

Axial rotation of the top cone

Having determined the static balancing conditions for rotations about any axis in the xy -plane, the only subsequent possible motion of the top cone relative to the bottom cone is a rotation about its' axis of revolution. It is important to verify whether or not the assumption that a pure rotation of the top cone about an axis in the xy -plane is possible without causing torsion instability about this axis of revolution. Using eqs.10) to 14), the total potential energy of the spring mechanism is determined with a rotation 2θ about the x -axis and a subsequent rotation ψ about the z' -axis of the top cone. The torque needed to establish the subsequent axial

rotation ψ of the top cone is defined as the derivative of the total potential energy to the rotation ψ :

$$M_{\psi} = \frac{\partial V_{tot}}{\partial \psi} \quad 19)$$

If the torque needed to establish the rotation ψ is positive, it is obvious that work has to be done to establish this subsequent rotation, which causes the potential energy of the mechanism to rise. The ‘statically balanced’ rotation of the top cone about an arbitrary axis in the xy -plane then is a ‘torsion stable’ one.

Having executed the matrix and vector operations symbolically with *Maple* it is again seen that the mechanism is only rotation symmetric (φ dependency eliminated) when $C_1 \cdot ab = C_2 \cdot cd = C_3 \cdot ef$ is substituted in the equation. The potential energy expression then reduces to:

$$V_{tot} = C_1 \cdot a \cdot b \cdot \left(-\frac{3}{2} \sin \beta \cos \psi \cos 2\theta \sin \alpha - \frac{3}{2} \sin \beta \cos \psi \sin \alpha \right) + K \quad 20)$$

With K the constant terms in the equation

The torque needed to establish the rotation ψ then becomes:

$$M_{\psi} = \frac{\partial V_{tot}}{\partial \psi} = \frac{3}{2} \cdot C_1 \cdot a \cdot b \cdot \sin \psi \cdot \sin \alpha \cdot \sin \beta \cdot (\cos 2\theta + 1) \quad 21)$$

It is clearly seen that a positive torque is needed to achieve the axial rotation ψ as long as $\sin \alpha \cdot \sin \beta \geq 0$. This torque decreases as 2θ increases until, at $2\theta=\pi$, the top cone is statically balanced for rotations ψ . Note that by rotating the top cone through an angle ψ at $2\theta=\pi$, the mechanism is basically the same as the 1dof spring mechanism shown in Fig.1-10, satisfying the conditions set out in eq.9). Furthermore, $\sin \alpha \sin \beta = 0$ when one of the apical angles is π and with one of the two apical angles negative, the mechanism becomes torsion unstable.

Summarising it has been shown that a 2dof mechanism with a basic configuration as given in Fig.1-12 can achieve perfect static balance for rotations about any axis lying in the xy -plane provided the conditions set out in eq.18) are satisfied. Taking into account the torsion stability of the mechanism however, one extra condition has to be satisfied:

$$\sin \alpha \cdot \sin \beta \geq 0 \quad 22)$$

Using the same reasoning set out in the previous section (Fig.1-8), the balancing conditions for the three spring balancer shown in Figs.1-12 and 13 (eq.18)) remain valid when the apexes of the cones of the mechanism are shifted apart or overlapped with a distance l_{sh} , provided springs with a free-length of $l_0=l_{sh}$ and $l_0=-l_{sh}$ respectively are employed and $\alpha=\beta$; $a=b$; $c=d$; and $e=f$. Figure 1-16 shows a possible embodiment of such a mechanism. Spheres with a radius r_s such that the free-length of the springs is $l_0=2r_s=l_{sh}$ can be used to ensure a correct rotation of top cone relative to the bottom cone with little hysteresis.

As is also the case for the 1dof mechanism, eq.18) is no longer valid when an extra spring or mass is subsequently mounted on the top cone (Also see Fig.1-11). Furthermore, because the cones are shifted apart/ overlapped, the torsion stiffness/ stability of the mechanism changes. Using a similar method as set out in this section, this can be looked into further should such a mechanism be applied.

Figure 1-16: Mechanism as Fig.1-12 with $a=b$; $c=d$; $e=f$; $\alpha=\beta$; and $\tan\alpha\tan\beta=2$. The apexes of the cones are shifted apart with a distance l_{sh} without changing the mechanisms characteristics, while springs with a free-length $l_0=l_{sh}$ are employed. Spheres with radius r_s are employed (such that $2r_s=l_{sh}=l_0$) to ensure little hysteresis. Note however, that a full 360° rotation will not be possible anymore.

1.2.3 Summary and Review

This section described several mechanisms that enable perfect static balance for rotations about a fixed pivot provided the determined balancing conditions are satisfied. This has been done for mechanisms containing both a 1dof pivot (hinge) as well as for 2dof pivot with a spherical joint. These basic mechanisms are summarized in table 1-1.

Several embodiments of these basic mechanisms are possible. A special embodiment that allowed for perfect static (spring-to-spring) balance by achieving ideal spring functionality with technical springs (Fig.1-8 and 1-16) has also been discussed. However, the basic mechanisms considered in this section have all been derived with ideal springs ($l_0=0$) between their attachment points. In conception, a perfect gravity balancing mechanism with technical springs applied directly between the spring attachment points is not possible with these mechanisms.

To achieve ideal spring functionality with technical springs (in practice), special constructions or springs are required (see Fig.1-5 and 1-6). Other methods of approximating static balance with technical springs include application of wrapping cams^{5, 7, 17} or non-linear springs using special coiling techniques. In short: Achieving ideal spring functionality with technical springs is possible but tends to complicate mechanism construction, especially with large spring pre-stresses in combination with relative large mechanism movements -as will also be the case for a possible ankle joint mechanism-. This is why none of the mechanisms found in this section seem suitable for further development into a prosthetic ankle joint, although theoretically possible.

<i>Mechanisms</i>	<i>Balancing conditions</i>
$C_3 hi = -(C_1 ab + C_2 cd) \cos(\alpha + \beta)$ <p style="text-align: center;">And</p> $C_4 hj = (C_1 ab - C_2 cd) \sin(\alpha + \beta)$

	$m_1 g R_1 = -(C_1 a b + C_2 c d) \cos(\alpha + \beta)$ <p style="text-align: center;">And</p> $m_2 g R_2 = (C_1 a b - C_2 c d) \sin(\alpha + \beta)$
	$C_1 \cdot a b = C_2 \cdot c d = C_3 \cdot e f$ <p style="text-align: center;">And</p> $C_s h i = -\frac{3}{2} C_1 a b \cos \alpha \cos \beta (2 - \tan \alpha \cdot \tan \beta)$ <p style="text-align: center;">And</p> $\sin \alpha \cdot \sin \beta \geq 0 \text{ (Torsion stability)}$
	$C_1 \cdot a b = C_2 \cdot c d = C_3 \cdot e f$ <p style="text-align: center;">And</p> $m g R = -\frac{3}{2} C_1 a b \cos \alpha \cos \beta (2 - \tan \alpha \cdot \tan \beta)$ <p style="text-align: center;">And</p> $\sin \alpha \cdot \sin \beta \geq 0 \text{ (Torsion stability)}$

Table 1-1: Overview of static balancing mechanisms with ideal linear tension springs between the spring attachment points, with: 2α or 2β the apical angles of a triangle or cone on which the lever arms for the spring attachment points a to i are mounted; R_i the lever arms on which the masses are mounted; C_i the spring stiffness.

1.3 Static gravity balancing with technical springs

The gravity balancing mechanisms described in the previous section, are derived with ideal springs ($l_0=0$) employed between their attachment points. However, the use of ideal springs in the conception phase leads to more difficult to construct mechanisms because the ideal spring functionality is generally –approximated- with technical springs. To facilitate construction of (gravity) balancers, employing technical springs ($l_0 \neq 0$) directly mounted between the link attachment points of the mechanism in the conception phase therefore seems beneficiary.

In the course of this section the design synthesis of such mechanisms will be explained and balancing conditions will be derived. Subsequently several approximate balancing concepts will be discussed. This is done, using the same methodology as set out in the previous sections.

1.3.1 1 dof static balancing

Consider the triangle/ spring mechanism given in Fig.1-17. The mechanism is as shown in Fig.1-7 but with the lever arms $a=b \neq c=d$. Furthermore the two employed linear springs with stiffness C_1 and C_2 also have a free-length of l_{01} and l_{02} respectively.

Figure 1-17: Mechanism as in Fig.1-7 but with the lever-arms $a=b \neq c=d$ and the employed springs have a stiffness C_1 and C_2 with a free-length l_{01} and l_{02} respectively.

Using the symbols shown in Fig.1-17, where it is especially noted that the upper half of the mechanism is rotated through an angle of 2θ , the following potential energy expression can be derived (see Appendix, section 1.6.2):

$$V = \cos 2\theta \cos 2\gamma (C_1 a^2 + C_2 b^2) + \sin 2\theta \sin 2\gamma (C_1 a^2 - C_2 b^2) - 2 \cos \theta \cos \gamma (C_1 l_{01} a + C_2 l_{02} b) - 2 \sin \theta \sin \gamma (C_1 l_{01} a - C_2 l_{02} b) + K \quad (23)$$

With $\gamma = \frac{1}{2}(\alpha + \beta)$ and K the constant terms in the expression.

From eq.23) it is readily seen that the potential energy of the mechanism is independent of the rotation 2θ of the top triangle when $C_1 a^2 = C_2 b^2$, $\alpha + \beta = 0.5\pi$ and $l_0 = 0$. The resulting mechanism then is the same as the static two-spring mechanism discussed in section 1.2.1 (Fig.1-7, eq.6)), only with $a = b \neq c = d$. If these static spring-to-spring balancing conditions are not satisfied a $\cos 2\theta$; $\sin 2\theta$; $\cos \theta$; and a $\sin \theta$ term times a constant remain in the expression. These terms can be eliminated from the expression with rotating masses using a similar approach as set out in the previous section. However, now two rotations (θ and 2θ) of the masses need to be coupled as shown in Fig.1-18.

Figure 1-18: An unstable mechanism consisting of three masses and two rollers. Mass m_1 and m_2 are connected to the top roller. Mass m_3 is connected to the connection line between the centres of the rollers.

Figure 1-18 shows mechanism consisting of three masses and two rollers with radius r_r . The rollers are connected such that only rolling of the top roller is possible relative to the bottom roller. This causes the connection line between the centres of the rollers to rotate through an angle θ for a rotation 2θ of the top roller. With the masses m_1 and m_2 mounted on the top roller and the mass m_3 on the -extension of- the connection line between the centers of the two rollers, the rotations are coupled correctly and the total potential energy expression of these masses for a rotation 2θ of the top roller becomes:

$$V_{mass} = \cos 2\theta \cdot g \cdot m_1 R_1 - \sin 2\theta \cdot g \cdot m_2 R_3 + \cos \theta \cdot g \cdot (m_1 R_2 + m_2 R_2 + m_3 R_4) + \sin \theta \cdot g \cdot m_3 R_5 \quad (24)$$

With R_2 twice the roller radius r_r .

With the potential energy expression of the mass mechanism containing the same products of θ -dependent goniometric terms and constants as for the spring mechanism, it should be possible to achieve perfect static balance by mounting the mass-mechanism to the spring-mechanism as shown in Fig.1-19. This is possible because the potential energy characteristic of the spring-mechanism given in Fig.1-17 is not influenced by shifting apart or overlapping the apexes with a distance $l_{sh}=2r_r$ provided the shifting distance is compensated for in the free-length of the -then technical- springs and $\alpha=\beta$ (Also review Fig.1-8).

Figure 1-19: The mechanisms given in Figs.1-17 and 1-18 combined. The triangle-spring mechanism is as described in Fig.1-17 only the apexes of the triangles are shifted apart with a distance $l_{sh}=2r_r$ and $\alpha=\beta$, according to Fig.1-8.

Figure 1-20: Mechanisms as given in Fig.1-19, but with: a) $m_2=0$ and $m_3=0$. b) $m_1=0$; $m_2=0$; $\alpha=0.25\pi$; and $R_5=0$.¹⁵ c) As b) but with triangles overlapped.^{14, 15} d) Working prototype of c), in the prototype, flexible bands are used to prevent the rollers from slipping.^{14, 15}

In combining the mass –and spring mechanism, the potential energy terms of both mechanisms can be summed to achieve the potential energy of the total mechanism and the following conditions for perfect static balance can be derived:

$$\begin{aligned} \cos 2\alpha(C_1 a^2 + C_2 b^2) &= -m_1 g R_1 \\ \sin 2\alpha(C_1 a^2 - C_2 b^2) &= m_2 g R_3 \\ 2\cos\alpha(C_1 l_{01} a + C_2 l_{02} b) &= g(m_1 R_2 + m_2 R_2 + m_3 R_4) \\ 2\sin\alpha(C_1 l_{01} a - C_2 l_{02} b) &= m_3 g R_5 \end{aligned} \tag{25}$$

Note that $\alpha=\beta$ thus $\gamma=\alpha=\beta$.

Figure 1-20 shows three possible ‘single mass’-spring mechanisms, directly derived from the previously described basic mechanism. It is noted that for eq.25) to remain valid, springs with free-length $l_{01}-l_{sh}$ and $l_{02}-l_{sh}$ have to be employed with overlapping the triangles (Fig.5-20c), springs with zero length $l_{01}+l_{sh}$ and $l_{02}+l_{sh}$ have to be employed with shifting them apart.

The basic mechanism given in Fig.1-19 allows for an infinite number of mass-spring combinations that enable perfect static balance. The basic thought is to zero out the cosine and sine terms in the potential energy expression of the spring-mechanism (eq.23)) by means of rotating masses or –ideal- springs. Since the goniometric terms in the potential energy expression contain both 2θ -and θ dependencies, a coupling -or combination of two rotations is then demanded to achieve this. This can be achieved with two rollers as described, but numerous different embodiments including gears, belts, or linkage type mechanisms can also be used. However, the use of rollers allows for a relative simple mechanism with little hysteresis during operation.

1.3.2 2 dof static balancing

Analogous to 1dof gravity balancing with technical springs, replacing the ideal springs in the 2dof spring balancer (Fig.1-12) with technical springs causes a residual torque that can be balanced with a mass rotating about two pivots (Fig.1-21).

The architecture of the spring-mechanism is basically the same as the 2dof spring balancer discussed in section 1.2.2 (Fig.1-12). Technical springs with stiffness C_1 , C_2 , and C_3 and free-length of l_{01} , l_{02} and l_{03} respectively, have replaced the three ideal springs. Furthermore the mechanism is now symmetric in that it is now equipped with three lever-arms $a \neq b \neq c$ and cones with equal apical angles 2α . With this set-up, the apexes of the cones can be shifted apart to be connected to the centres of two spheres without changing the spring-mechanisms characteristics, provided the shifting distance l_{sh} is compensated for in the free-length of the employed springs as shown in Fig.1-21 (also see Fig.1-8). A mass m can then be mounted directly to the axis of revolution of the top cone.

Figure 1-21: Conception of 2dof gravity balancer consisting of 6 levers, three technical springs and two cones, interconnected with a spherical joint in the apexes.

Using a similar approach as set out in the previous section, the conditions for perfect static balance can now be derived for rotations of the mass relative to the neutral position. Subsequently the torsion stability of each ‘balancing position’ can be considered.

The potential energy expression of the three-spring mechanism (Fig.1-21b) for a rotation 2θ of the top cone relative to the bottom cone is (see Appendix section 1.6.3):

$$V_{spring} = 1.5 \cdot C_1 a^2 \cos^2 \alpha (2 - \tan^2 \alpha) \cos 2\theta - 6 \cdot C_1 \cdot l_{01} \cdot a \cdot \cos \alpha \cdot \cos \theta + K \quad (26)$$

With K the constant terms; $C_1 \cdot a^2 = C_2 \cdot b^2 = C_3 \cdot c^2$; $C_1 \cdot a \cdot l_{01} = C_2 \cdot b \cdot l_{02} = C_3 \cdot c \cdot l_{03}$; and $R_2 = 2r_s$.

Where the potential energy expression of the mass (Fig.1-21a) is:

$$V_{mass} = mgR_1 \cdot \cos 2\theta + mgR_2 \cdot \cos \theta + K \quad (27)$$

The total potential energy expression for the total mechanism (Fig.1-21c) then becomes:

$$V_{tot} = V_{spring} + V_{mass} = 1.5 \cdot C_1 a^2 \cos^2 \alpha (2 - \tan^2 \alpha) \cos 2\theta - 6 \cdot C_1 \cdot l_{01} \cdot a \cdot \cos \alpha \cdot \cos \theta + mgR_1 \cdot \cos 2\theta + mgR_2 \cdot \cos \theta + K \quad (28)$$

With K the constant terms; $C_1 \cdot a^2 = C_2 \cdot b^2 = C_3 \cdot c^2$; $C_1 \cdot a \cdot l_{01} = C_2 \cdot b \cdot l_{02} = C_3 \cdot c \cdot l_{03}$; and $R_2 = 2r_s$.

Equation 28) yields the following conditions for perfect gravity balance of the mechanism given in Fig.1-21c:

$$\begin{aligned} mgR_1 &= -1 \frac{1}{2} C_1 a^2 \cos^2 \alpha (2 - \tan^2 \alpha) \\ mgR_2 &= 6 l_{01} C_1 a \cos \alpha \end{aligned} \quad (29)$$

With: $C_1 \cdot a^2 = C_2 \cdot b^2 = C_3 \cdot c^2$, $C_1 \cdot a \cdot l_{01} = C_2 \cdot b \cdot l_{02} = C_3 \cdot c \cdot l_{03}$ and $R_2 = 2r_s$.

With the static balancing conditions known, the torsion stability conditions or stiffness of this mechanism can be determined, using a potential energy consideration. Because however this was done numerically (not symbolically as was the case for the 2dof balancer with ideal springs) and there are many –geometric- system variables, only the method is set out in the Appendix (section 1.6.4). This should be looked into further if such a mechanism is actually going to be employed and torsion-stability or rather torsion-stiffness of the mechanism is an issue.

Figure 1-22: Working prototype of the mechanism given in Fig.1-21. All lever arms and springs are equal.

Figure 1-22 shows a working prototype of the basic mechanism displayed in Fig.1-21. It should be noted that the spheres rely only on friction to ensure a 2dof movement. This is because flexible bands could not be used, as was the case for the 1dof mechanism. The prototype incorporates six equal lever arms and three equal linear technical springs. Finally, the mechanism proved torsion stable in theory and –more important- in practice.

Although the prototype worked quite satisfactory when the correct balancing conditions were set, during experiments it is observed that a correct alignment is as necessary as it is difficult to obtain and sustain. This is of course because of the lack of hysteresis with the current rolling link embodiment. Although such an embodiment is theoretically possible, it is too difficult to construct and too unreliable for further development into a prosthetic ankle joint.

1.3.3 Static balancing with compression springs

The previous sections all discussed balancing mechanisms incorporating linear technical tension springs. Sometimes however the use of compression springs has advantages over tension springs. Especially when large spring forces need to be transmitted within a compact mechanism the use of compression springs can be useful.

Considering the basic one -or 2dof spring-to-spring mechanisms incorporating linear technical tension springs, the potential energy expression has consequently been determined as the sum of the energy stored in each spring, thus:

$$\frac{1}{2}C_i u_i^2 = \frac{1}{2}C_i (l_i - l_{0i})^2 = \frac{1}{2}C_i (l_i^2 + l_{0i}^2 - 2l_i l_{0i})$$

With l_i the distance between attachment points, l_{0i} the zero length; and C_i the stiffness of spring i .

Employing linear compression springs in these mechanisms gives the following expression for the energy stored in each spring:

$$\frac{1}{2}C_i u_i^2 = \frac{1}{2}C_i (l_{0i} - l_i)^2 = \frac{1}{2}C_i (l_i^2 + l_{0i}^2 - 2l_i l_{0i})$$

As can be seen the spring energy expressions are the same for both the tension as the compression springs meaning that the balancing conditions determined for the mechanisms employing technical tension springs are also valid for compression springs. Note however that simply applying compression springs as if they were tension springs in 2dof mechanisms can cause torsion instabilities.

1.3.4 Approximate gravity balancing with technical springs

This chapter has described several perfect gravity balancing concepts. Mechanisms have been discussed that incorporated ideal springs, but also mechanisms in which technical springs were employed have been discussed. To obtain ideal spring functionality with technical springs requires special constructions in practice, where employing technical springs directly between the links demands a coupling of two rotations. Although theoretically and physically possible, these constructional demands make the architecture of these mechanisms too complex for a future prosthetic ankle joint.

This section discusses several easy to construct gravity balancing mechanisms that approximate the perfect balance extremely well with deviations from perfect balance between 0.6% and 1.5% within the operation range of a future ankle joint. Because these mechanisms are mechanically simple, they do seem suitable for further development into a prosthetic ankle joint. The fact that approximate balance can only be achieved, is acceptable since the goal of

the gravity balancing prosthetic ankle joint is to estimate the initial ankle joint stiffness required for normal walking. It is as unlikely that a prosthetic user will notice the 1.5% deviation in ankle joint stiffness as it is unlikely that the perfect gravity balancer will function as such during operation (walking), where ease of construction and simplicity are insuperable demands in prosthetic engineering.

First the discussed gravity balancers employed with technical springs are considered for small angles of rotation, which is reasonable since the maximum occurring rotations during operation are approximately 10° . Secondly the deviation from perfect balance will be discussed when the gravity of a mass rotating about a fixed pivot is approximately balanced with one technical spring.

1dof gravity balance for small angles of rotation

Figure 1-23: a) A 1dof gravity balancer for small angles of rotation. b) Working prototype of a). Rolling links are used to minimize hysteresis during operation.

The mechanism shown in Fig.1-23 consists of the spring-mechanism shown in Fig.1-17 with a mass mounted on the top triangle. Assuming that the conditions $C_1 a^2 = C_2 b^2$ and $C_1 l_{01} a = C_2 l_{02} b$ of the spring mechanism are satisfied, the total potential energy expression becomes (also see eq.23)):

$$V_{tot} = V_s + V_m = 2C_1 a^2 \cdot \cos 2\theta \cdot \cos 2\gamma - 4C_1 l_{01} a \cdot \cos \theta \cdot \cos \gamma + mgR \cdot \cos 2\theta + K \quad 30)$$

With $\gamma = \frac{1}{2}(\alpha + \beta)$; $C_1 a^2 = C_2 b^2$; $C_1 l_{01} a = C_2 l_{02} b$ and K the constant terms.

Considering this expression contains both θ and 2θ dependencies it is no longer possible to achieve perfect gravity balance. However, it remains possible to derive gravity-balancing conditions for small angles of rotation from the neutral position of the top triangle ($2\theta=0$). As discussed in the introduction it is then necessary for the first derivative of the potential energy expression to the rotation angle 2θ to be zero. Subsequently the second derivative has to be positive. If the second derivative is zero, then the third has to be zero and fourth has to be positive, and so on.

Taking the first derivative to the rotation 2θ of the top triangle relative to the bottom triangle gives the torque needed to achieve that rotation:

$$V_{tot}' = \frac{\partial V_{tot}}{\partial 2\theta} = -2C_1a^2 \cos 2\gamma \sin 2\theta + 2C_1l_{01}a \cos \gamma \sin \theta - mgR \sin 2\theta \quad 31)$$

By subsequently considering the mechanism for small angles of rotation the following balancing condition can be derived:

$$\begin{aligned} V_{tot}' &= -4C_1a^2 \cdot \cos 2\gamma \cdot \theta + 2C_1l_{01}a \cdot \cos \gamma \cdot \theta - 2mgR \cdot \theta \Rightarrow \\ &-4C_1a^2 \cdot \cos 2\gamma + 2C_1l_{01}a \cdot \cos \gamma - 2mgR = 0 \end{aligned} \quad 32)$$

$$\text{With } \gamma = \frac{1}{2}(\alpha + \beta); \quad C_1a^2 = C_2b^2; \quad \text{and} \quad C_1l_{01}a = C_2l_{02}b.$$

Assuming this balancing condition is satisfied, the second and third derivative are equal to zero. With the fourth derivative being always greater than zero ($\alpha+\beta<\pi$), it is seen that the mechanism is slightly overbalanced. Considering a maximum angular rotation $2\theta=10^\circ$ of the top triangle, a maximum deviation from perfect balance of a mere 0.6% is revealed after substituting the balancing condition back into eq.31).

The main advantage of this mechanism is the relative ease in construction when compared to the discussed perfect balancers since a kinematic coupling of two different segments is no longer needed. Figure 1-23b shows a working prototype of a mechanism satisfying the conditions set out in eq.32). The prototype consists of two –virtual- triangles with equal apical angles and lever-arms. Also the mechanism is equipped with rollers to minimise hysteresis during operation (also see Fig.1-8). The apexes of the –virtual- triangles are slightly overlapped which has a slightly stabilizing effect but makes the mechanism self-adjusting and therefore easy to set.

In practice this model functions surprisingly well. Within the operation range, the mass can be rotated in every position and remain in that position. Considering hysteresis is minimal during operation, this suggests a correct working of the mechanism.

2dof gravity balance for small angles of rotation

Using a similar approach as for the 1dof gravity balancing mechanism with technical springs, the 2dof balancing mechanism with technical springs can be considered for small angles of rotation. In this case a mass is mounted directly to the axis of revolution of the top cone of the three-spring-mechanism shown in Fig.1-21b. The total potential energy expression then becomes (Also see eq.26)):

$$V = 1.5 \cdot C_1 a^2 \cos^2 \alpha (2 - \tan^2 \alpha) \cos 2\theta - 6 \cdot C_1 l_{01} a \cos \alpha \cdot \cos \theta + mgR \cdot \cos 2\theta + K \quad (33)$$

With K the constant terms; $C_1 a^2 = C_2 b^2 = C_3 c^2$ and $C_1 a l_{01} = C_1 b l_{02} = C_3 c l_{03}$

Taking the first derivative to the rotation 2θ of the top cone relative to the bottom cone gives the torque needed to achieve that rotation:

$$\begin{aligned} V_{tot}' = \frac{\partial V_{tot}}{\partial 2\theta} = & -1.5 \cdot C_1 a^2 \cdot \cos^2 \alpha \cdot (2 - \tan^2 \alpha) \sin 2\theta \\ & + 3 \cdot C_1 l_{01} a \cdot \cos \alpha \cdot \sin \theta - mgR \cdot \sin 2\theta \end{aligned} \quad (34)$$

By subsequently considering the mechanism for small angles of rotation the following balancing condition can be derived:

$$\begin{aligned} V_{tot}' = & -3C_1 a^2 \cdot \cos^2 \alpha \cdot (2 - \tan^2 \alpha) \cdot \theta + 3C_1 l_{01} a \cdot \cos \alpha \cdot \theta - 2mgR \theta \Rightarrow \\ & -3C_1 a^2 \cdot \cos^2 \alpha \cdot (2 - \tan^2 \alpha) + 3C_1 l_{01} a \cdot \cos \alpha - 2mgR = 0 \end{aligned} \quad (35)$$

With $C_1 a^2 = C_2 b^2 = C_3 c^2$ and $C_1 a l_{01} = C_2 b l_{02} = C_3 c l_{03}$

Assuming this balancing condition is satisfied, it is seen that the second and third derivative are equal to zero and the fourth derivative is always greater than zero ($\alpha < 0.5\pi$). Meaning that this mechanism is slightly overbalanced –as was also the case for the 1dof mechanism. Finally, considering a maximum angular rotation $2\theta = 10^\circ$ of the top triangle, a maximum deviation from perfect balance of a mere 0.6% is also revealed after substituting the balancing condition back into eq.34).

By replacing the ideal springs with technical springs, the torsion stiffness or stability conditions of this mechanism can be determined using a similar approach as set out in section 1.2.2, should this be an issue in a future –ankle- design (see Appendix, section 1.6.4).

As was also the case for the 1dof gravity balancer, the main advantage of this 2dof mechanism is the relative ease in construction when compared to the discussed perfect 2dof balancers since a kinematic coupling of two different rotating segments is no longer needed. Figure 1-24b shows a working prototype of a mechanism satisfying the conditions set out in eq.35). The prototype consists of two cones with equal apical angles and lever-arms. Also the mechanism

is equipped with a ‘sharp’ ball-and-socket joint to minimize hysteresis during operation (Also see Fig.1-8).

In practice this model also functions surprisingly well. Meaning the mass can be rotated in every position and remain in that position for rotations well over the intended operation range (without torsion instability).

Figure 1-24: a) 2dof gravity balancer for small angles of rotation. b) Working model of a). Rolling links are used to minimize hysteresis during operation.

Approximate balance with a single technical spring and mass

This paragraph discusses a mechanism consisting of a single mass rotating about a fixed pivot of which gravity is approximately balanced with a single technical spring. The technical spring has a free-length l_0 (Fig.1-25). First the spring characteristic required for perfect static balance is determined. Subsequently the deviation from perfect balance is determined when a simple linear technical spring is employed.

The required spring-force to achieve perfect gravity balance of the mass m is:

$$F_{s_req} = \frac{mgR \sin \theta}{b \sin \beta} = \frac{mgR \sin \theta \cdot l_1}{ab \sin(\varphi_0 + \theta)} = K l_1 \cdot \frac{l_1 \cdot \sin \theta}{\sin(\varphi_0 + \theta)} \quad 36)$$

Figure 1-25: A mass rotating about a fixed pivot that is approximately balanced with one linear technical spring (free-length l_0). a) Mechanism in neutral position with spring relaxed. b) Mass rotated through an angle θ . c) As b) with spring force F_s and gravitational force mg acting on the mass displayed.

The spring-force exerted for a rotation θ of link R when assuming a linear technical spring is:

$$F_s = C_s \cdot \left(\sqrt{a^2 + b^2 - 2ab \cos(\varphi_0 + \theta)} - \sqrt{a^2 + b^2 - 2ab \cos \varphi_0} \right) = K_2 + C_s \cdot l_1 \quad (37)$$

With C_s the spring stiffness.

It is seen that eq.36) and eq.37) are never equal, also when only small angles of rotation are considered (except when $\varphi_0 = 0$ and $a = b$, in which case an ideal spring is employed and the condition set out in eq.2) is satisfied). To assess the deviation from perfect balance, the required (linear) spring stiffness is estimated by assuming that the spring exerts the required force at the maximum rotation of the mass (10°). Subsequently the spring-force between 0 and 10° is considered linear.

Thus the spring stiffness is determined according to:

$$C_s = \frac{F_{s_req_10^\circ}}{l_1 - l_0} = \frac{F_{s_req_10^\circ}}{\sqrt{a^2 + b^2 - 2ab \cos(\varphi_0 + \theta)} - \sqrt{a^2 + b^2 - 2ab \cos \varphi_0}} \quad (38)$$

With the spring stiffness determined according to eq.38), the counteracting torque of the spring for a rotation θ of the mass is:

$$M_s = C_s \cdot (l_1 - l_0) \cdot b \cdot \sin \beta \quad (39)$$

Where the required torque of static balance the rotating mass is:

$$M_{req.} = mgR \sin \theta \quad 40)$$

With the mechanism dimensions according to the dimensions of the prototype prosthetic foot discussed in the previous chapter ($a=50\text{mm}$, $b=140\text{mm}$, $\phi_0=62^\circ$), the maximum deviation from perfect static balance is 1.5% during operation. Of course the main advantage of this approximate balancing mechanism are the relative ease of construction and simple design.

1.3.5 Summary and Review

Several mechanisms that statically balance gravity of a mass rotating about a fixed pivot, with technical (nonzero free-length) springs employed directly between the spring attachment points, have been described. Both conditions for perfect as well as for balancing for small angles of rotations have been derived. This has been done for mechanisms employed with both a 1dof pivot (hinge) as well as for a 3dof pivot (spherical joint) of which the basic configurations are summarized in tables 1-2 and 1-3 together with the balancing conditions.

It been shown that perfect gravity balancing is possible when the ideal springs -of the basic ideal spring balancer configuration- are replaced with technical springs. However, to obtain this perfect balance with technical springs, two rotating segments always have to be coupled such that a rotation of 2θ of one segment causes a rotation θ of the other segment. This kinematic demand can be satisfied by stacking two rollers (1dof) -or spheres (2dof) in which a rotation 2θ of one roller -or sphere relative to the other, causes a rotation θ of the connection line between the centres of the rollers (assuming pure rolling). However, although this kinematic coupling of two rotating segments results in mechanisms that are relative easy to construct (compared to other mechanisms employing technical springs), these mechanisms remain too difficult to construct when the aim is to use them as future prosthetic ankle joints. In addition to that, experiments reveal that balancing conditions are difficult to set and sustain with two stacked spheres, relying only on friction to prevent slipping. This makes –this embodiment of- the 2dof balancer too unreliable for a future prosthetic ankle joint.

Besides perfect balancing, approximate gravity balancing concepts also have been discussed. Instead of coupling the rotations of two different segments to establish perfect gravity balance with technical springs, the mechanisms are considered for small angles of rotation. This is reasonable since during operation, angles of 10° maximum are approximately established. Under these conditions, approximate gravity balancing with a 99.4% similarity with perfect gravity balancing is possible (for 1dof as well as for the 2dof mechanisms). The prototypes resulting from this approach are mechanically simple and function surprisingly well during operation, making them excellent candidates for static gravity balancing purposes. The final

part of the approximate balancing section discussed a 1dof balancing mechanism in which a rotating mass is approximately balanced with one linear technical spring. Although perfect –or even approximate for small angles of rotation- balancing is not possible with this mechanism, the maximum deviation from perfect static balancing, within the operation range of a prosthetic ankle joint, is a mere 1.5%. This relative small deviation from perfect static balance, in combination with the relative ease of construction, makes this mechanism very suitable for further development into a –1dof - prosthetic ankle joint.

Perfect 2dof gravity balance with technical springs	
Balancing conditions:	
$mgR_1 = -1/2 C_1 a^2 \cos^2 \alpha (2 - \tan^2 \alpha)$ $mgR_2 = 6l_{01} C_1 a \cos \alpha$ $C_1 \cdot a^2 = C_2 \cdot b^2 = C_3 \cdot c^2$ $C_1 \cdot a \cdot l_{01} = C_2 \cdot b \cdot l_{02} = C_3 \cdot c \cdot l_{03}$	
Torsion stiffness is mechanism dependent: See appendix	

Table 1-2: Overview of perfect static gravity balancing mechanisms with technical springs between the spring attachment points, with: 2α or 2β the apical angles of a triangle or cone on which the lever arms for the spring attachment points a to f are mounted; R_i the lever arms on which the masses are mounted; l_{0i} the spring free-length; C_i the spring stiffness.

Gravity balancing for small angles of rotation	
<p>1dof</p>	$-4C_1a^2 \cdot \cos(\alpha + \beta) + 2C_1l_{01}a \cdot \cos \frac{1}{2}(\alpha + \beta) - 2mgR = 0$ $C_1a^2 = C_2b^2$ $C_1l_{01}a = C_2l_{02}b$
<p>2dof</p>	$-3C_1a^2 \cdot \cos^2 \alpha \cdot (2 - \tan^2 \alpha) + 3C_1l_{01}a \cdot \cos \alpha - 2mgR = 0$ $C_1a^2 = C_2b^2 = C_3c^2$ $C_1al_{01} = C_2bl_{02} = C_3cl_{03}$ <p><i>Torsion stiffness is mechanism dependent: See appendix (section 1.6.4)</i></p>

Table 1-3: Overview of mechanisms that statically balance gravity for small angles of rotation with: 2α or 2β the apical angles of a triangle or cone on which the lever arms for the spring attachment points a to f are mounted; R the lever arms on which the masses are mounted; l_{0i} the spring free-length; C_i the spring stiffness.

1.4 Concluding remarks

This section has shown several embodiments (1dof –as well as 2dof) of gravity balancers incorporating ideal springs. The difficulty of ideal springs is that special constructions are generally needed, to obtain ideal spring functionality with technical springs, making these mechanisms difficult to construct and too complex for further development into a prosthetic ankle joint.

It has also been show that perfect gravity balancing can be established with technical springs employed directly between the links of a mechanism. However, to obtain perfect balance with technical springs, two rotating segments always have to be coupled such that a rotation of 2θ of one segment causes a rotation θ of the other segment. In this chapter this kinematic condition is satisfied by means of two stacked rollers (1dof) or spheres (3dof) in which a rotation 2θ of one roller or sphere relative to the other (assuming pure rolling), causes a rotation θ of the connection line between the centres of the rollers or spheres. However, especially for 3dof embodiment, construction remains too difficult for a future prosthetic ankle application.

The final section of this part of the thesis discussed some mechanically simple approximate gravity balancing mechanisms with up to 99.4% similarity with perfect gravity balancing within the operation range ($\sim 10^\circ$ rotation of mass). Working prototypes of the mechanisms have been made that functioned quite satisfactory, well beyond the hypothesized operation range.

Considering that the goal is to incorporate the balancing mechanism in the future prosthetic foot, simplicity in construction is an insuperable demand where there is no strict demand for perfect balance. After all, static balancing of gravity acting on the center of gravity of a subject will and can only be used as a guideline to set the initial ankle joint stiffness. Furthermore it is not likely that a prosthetic walker will notice a 2-5% deviation in ankle joint stiffness. Therefore -since a mechanically simple perfect balancer has not been found- approximate balancing will of course also suffice.

The main goal of the prototype foot being to assess the effect of foot plantarflexion on ambulation during clinical trials, it was decided that –initially- only the 1dof ankle joint would be incorporated in the prototype foot. This is mainly to allow for a relative easy to construct prototype foot, but also to narrow down the number of ‘unknowns’ during the first clinical trials. For that purpose, the mechanism shown in Fig.1-25 seems the most suitable: It is an extremely easy to construct mechanism with an approximate 98.5% similarity with perfect balancing.

1.5 References

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1.6 Appendix

1.6.1 Potential energy 2dof ideal spring mechanism (1.2.2)

Besides using vector notation and calculation to determine the static balancing conditions for rotations of the top cone about an axis in the xy -plane, this can also be derived directly from the projection of the spring lengths on the zy -plane as described in section 1.2.2 (Fig.1-14).

Figure 1-26: Mechanism as described in section 1.2.2, projected on the xy –and yz -plane. a^* to f^* are the projections of the levers a to f on the yz -plane respectively. a) The mechanism in neutral position. b) Top cone of the mechanism rotated through an angle of 2θ about the x -axis.

Using the symbols from Fig.1-26, the projected spring-lengths l_1 to l_3 can be determined from the cosine-rule:

$$\begin{aligned}
 l_1^2 &= a^{*2} + b^{*2} - 2a^*b^* \cdot \cos\psi_1 \\
 l_2^2 &= c^{*2} + d^{*2} - 2c^*d^* \cdot \cos\psi_2 \\
 l_3^2 &= e^{*2} + f^{*2} - 2e^*f^* \cdot \cos\psi_3
 \end{aligned} \tag{1}$$

With $\psi_1 = \pi - \rho_1 - \rho_2 + 2\theta$; $\psi_2 = \pi - \rho_3 - \rho_4 - 2\theta$; $\psi_3 = \pi - \rho_5 - \rho_6 - 2\theta$; and a^* to f^* the lever-arm lengths a to f , projected on the yz -plane.

The cosine terms in eq.1) can be written as:

$$\begin{aligned}
 \cos(\psi_1) &= -\cos 2\theta \cos(\rho_1 + \rho_2) - \sin 2\theta \sin(\rho_1 + \rho_2) \\
 \cos(\psi_2) &= -\cos 2\theta \cos(\rho_3 + \rho_4) + \sin 2\theta \sin(\rho_3 + \rho_4) \\
 \cos(\psi_3) &= -\cos 2\theta \cos(\rho_5 + \rho_6) + \sin 2\theta \sin(\rho_5 + \rho_6)
 \end{aligned} \tag{2}$$

Writing out further gives:

$$\begin{aligned}
 \cos(\psi_1) &= -\cos 2\theta (\cos \rho_1 \cos \rho_2 - \sin \rho_1 \sin \rho_2) - \sin 2\theta (\sin \rho_1 \cos \rho_2 + \cos \rho_1 \sin \rho_2) \\
 \cos(\psi_2) &= -\cos 2\theta (\cos \rho_3 \cos \rho_4 - \sin \rho_3 \sin \rho_4) + \sin 2\theta (\sin \rho_3 \cos \rho_4 + \cos \rho_3 \sin \rho_4) \\
 \cos(\psi_3) &= -\cos 2\theta (\cos \rho_5 \cos \rho_6 - \sin \rho_5 \sin \rho_6) + \sin 2\theta (\sin \rho_5 \cos \rho_6 + \cos \rho_5 \sin \rho_6)
 \end{aligned}$$

Finally the angles ρ_i can be derived from:

$$\begin{aligned}
 \sin(\rho_1) &= \frac{a \cdot \sin(\alpha) \cdot \cos(\varphi)}{a^*}; \sin(\rho_2) = \frac{b \cdot \sin(\beta) \cdot \cos(\varphi)}{b^*}; \cos(\rho_1) = \frac{a \cdot \cos(\alpha)}{a^*}; \cos(\rho_2) = \frac{b \cdot \cos(\beta)}{b^*} \\
 \sin(\rho_3) &= \frac{c \cdot \sin(\alpha) \cdot \cos(\frac{1}{3}\pi - \varphi)}{c^*}; \sin(\rho_4) = \frac{d \sin(\beta) \cos(\frac{1}{3}\pi - \varphi)}{d^*}; \cos(\rho_3) = \frac{c \cdot \cos(\alpha)}{c^*}; \cos(\rho_4) = \frac{d \cdot \cos(\beta)}{d^*} \\
 \sin(\rho_5) &= \frac{e \cdot \sin(\alpha) \cdot \cos(\frac{1}{3}\pi + \varphi)}{e^*}; \sin(\rho_6) = \frac{f \sin(\beta) \cos(\frac{1}{3}\pi + \varphi)}{f^*}; \cos(\rho_5) = \frac{e \cdot \cos(\alpha)}{e^*}; \cos(\rho_6) = \frac{f \cdot \cos(\beta)}{f^*}
 \end{aligned} \tag{3}$$

Substitution of eq.2) and eq.3) in eq.1) gives:

$$\begin{aligned}
 l_1^2 &= a^{*2} + b^{*2} - 2ab \begin{pmatrix} -\cos 2\theta \cos \alpha \cos \beta + \cos 2\theta \sin \alpha \sin \beta \cos^2 \varphi - \sin 2\theta \sin \alpha \cos \beta \cos \varphi \\ -\sin 2\theta \cos \alpha \sin \beta \cos \varphi \end{pmatrix} \\
 l_2^2 &= c^{*2} + d^{*2} - 2cd \begin{pmatrix} -\cos 2\theta \cos \alpha \cos \beta + \cos 2\theta \sin \alpha \sin \beta \cos^2 (\frac{1}{3}\pi - \varphi) + \sin 2\theta \sin \alpha \cos \beta \cos (\frac{1}{3}\pi - \varphi) \\ + \sin 2\theta \cos \alpha \sin \beta \cos (\frac{1}{3}\pi - \varphi) \end{pmatrix} \\
 l_3^2 &= e^{*2} + f^{*2} - 2ef \begin{pmatrix} -\cos 2\theta \cos \alpha \cos \beta + \cos 2\theta \sin \alpha \sin \beta \cos^2 (\frac{1}{3}\pi + \varphi) + \sin 2\theta \sin \alpha \cos \beta \cos (\frac{1}{3}\pi + \varphi) \\ + \sin 2\theta \cos \alpha \sin \beta \cos (\frac{1}{3}\pi + \varphi) \end{pmatrix}
 \end{aligned} \tag{4}$$

With these geometric relations determined, the potential energy terms that vary for rotations 2θ of the top cone total can be determined according to:

$$V = \frac{1}{2}C_1 \cdot l_1^2 + \frac{1}{2}C_2 \cdot l_2^2 + \frac{1}{2}C_3 \cdot l_3^2 + K; \tag{5}$$

With K the constant terms and l_1 to l_3 the spring lengths projected on the yz -plane.

Substituting eq.4) into eq.5) gives:

$$\begin{aligned}
 V = & -C_1 ab \left(\begin{aligned} & -\cos 2\theta \cos \alpha \cos \beta + \cos 2\theta \sin \alpha \sin \beta \cos^2 \varphi - \sin 2\theta \sin \alpha \cos \beta \cos \varphi \\ & -\sin 2\theta \cos \alpha \sin \beta \cos \varphi \end{aligned} \right) \\
 & - C_2 cd \left(\begin{aligned} & -\cos 2\theta \cos \alpha \cos \beta + \cos 2\theta \sin \alpha \sin \beta \cos^2 (\frac{1}{3}\pi - \varphi) + \sin 2\theta \sin \alpha \cos \beta \cos (\frac{1}{3}\pi - \varphi) \\ & + \sin 2\theta \cos \alpha \sin \beta \cos (\frac{1}{3}\pi - \varphi) \end{aligned} \right) \\
 & - C_3 ef \left(\begin{aligned} & -\cos 2\theta \cos \alpha \cos \beta + \cos 2\theta \sin \alpha \sin \beta \cos^2 (\frac{1}{3}\pi + \varphi) + \sin 2\theta \sin \alpha \cos \beta \cos (\frac{1}{3}\pi + \varphi) \\ & + \sin 2\theta \cos \alpha \sin \beta \cos (\frac{1}{3}\pi + \varphi) \end{aligned} \right) + K
 \end{aligned} \tag{6}$$

With K the constant terms in the equation

Knowing that:

$$\begin{aligned}
 \cos^2 (\frac{1}{3}\pi - \varphi) + \cos^2 (\frac{1}{3}\pi + \varphi) &= \frac{1}{2} \cos^2 \varphi + 1.5 \sin^2 \varphi \\
 \cos (\frac{1}{3}\pi - \varphi) + \cos (\frac{1}{3}\pi + \varphi) &= \cos \varphi
 \end{aligned}$$

It is seen that the φ -dependency can only be eliminated from the eq.6) when $C_1 \cdot ab = C_2 \cdot cd = C_3 \cdot ef$. In other words: Then the mechanism is rotation symmetric. In doing so the potential energy equation becomes:

$$V = -C_1 ab (-3 \cos 2\theta \cos \alpha \cos \beta + 1 \frac{1}{2} \cdot \cos 2\theta \sin \alpha \sin \beta) + K ; \tag{7}$$

Thus:

$$V = -1 \frac{1}{2} \cdot C_1 ab \cos 2\theta (-2 \cos \alpha \cos \beta + \sin \alpha \sin \beta) + K ;$$

Thus:

$$V = -1 \frac{1}{2} \cdot C_1 ab \cos \alpha \cos \beta \cos 2\theta (-2 + \tan \alpha \tan \beta) + K ;$$

Now it is easily seen that the potential energy remains constant for rotations 2θ of the top cone if:

$$\tan \alpha \cdot \tan \beta = 2 \text{ and } C_1 \cdot ab = C_2 \cdot cd = C_3 \cdot ef . \tag{8}$$

1.6.2 Potential energy 1dof technical spring mechanism (1.3.1)

Figure 1-27: 1dof mechanism with technical springs employed between the spring attachment points on the lever arms a and b . The springs employed have a stiffness C_1 and C_2 with a zero length l_{01} and l_{02} respectively.

In the course of this section the total potential energy expression of the mechanism shown in Fig.1-27 is determined. The mechanism consists of two triangles with apical angles of 2α and 2β , interconnected with a hinge joint at the apex. The lever-arm lengths are a and b . Furthermore two linear tension springs are employed with stiffness C_1 and C_2 and a free-length of l_{01} and l_{02} respectively. Using the symbols shown in Fig.1-27, the following geometrical relations can be identified:

$$\psi_1 = \frac{1}{2}\pi - \frac{1}{2}(\alpha + \beta) + \theta; \quad \psi_2 = \frac{1}{2}\pi - \frac{1}{2}(\alpha + \beta) - \theta; \quad \ell_1 = 2a \sin \psi_1; \quad \ell_2 = 2a \sin \psi_2 \quad 9)$$

Writing out eq.9) gives:

$$l_1 = 2a \sin(\psi_1) = 2a \cos(\gamma - \theta) = 2a(\cos \gamma \cos \theta + \sin \gamma \sin \theta) \quad 10)$$

$$l_2 = 2b \sin(\psi_2) = 2b \cos(\gamma + \theta) = 2b(\cos \gamma \cos \theta - \sin \gamma \sin \theta)$$

Thus:

$$l_1^2 = 4a^2(\cos^2 \gamma \cos^2 \theta + \sin^2 \gamma \sin^2 \theta + 2 \cos \gamma \cos \theta \sin \gamma \sin \theta)$$

$$l_2^2 = 4b^2(\cos^2 \gamma \cos^2 \theta + \sin^2 \gamma \sin^2 \theta - 2 \cos \gamma \cos \theta \sin \gamma \sin \theta)$$

$$\text{With } \gamma = \frac{1}{2}(\alpha + \beta)$$

The total potential energy of this mechanism yields:

$$V = \frac{1}{2}C_1(l_1^2 - 2l_{01}l_1 + l_{01}^2) + \frac{1}{2}C_2(l_2^2 - 2l_{02}l_2 + l_{02}^2) \quad 11)$$

With substitution of eq.10) in eq.11) the total potential energy of the mechanism becomes:

$$\begin{aligned} V = & \frac{1}{2}C_1 \cdot \left(4a^2(\cos^2 \gamma \cos^2 \theta + \sin^2 \gamma \sin^2 \theta + 2 \cos \gamma \cos \theta \sin \gamma \sin \theta) \right. \\ & \left. - 4l_{01}a(\cos \gamma \cos \theta + \sin \gamma \sin \theta) + l_{01}^2 \right) \\ & + \frac{1}{2}C_2 \cdot \left(4b^2(\cos^2 \gamma \cos^2 \theta + \sin^2 \gamma \sin^2 \theta - 2 \cos \gamma \cos \theta \sin \gamma \sin \theta) \right. \\ & \left. - 4l_{02}b(\cos \gamma \cos \theta - \sin \gamma \sin \theta) + l_{02}^2 \right) \end{aligned}$$

Rewriting gives:

$$\begin{aligned} V = & \cos 2\theta \cos 2\gamma (C_1 a^2 + C_2 b^2) + \sin 2\theta \sin 2\gamma (C_1 a^2 - C_2 b^2) \\ & - 2 \cos \theta \cos \gamma (C_1 l_{01} a + C_2 l_{02} b) - 2 \sin \theta \sin \gamma (C_1 l_{01} a - C_2 l_{02} b) + K \end{aligned} \quad 12)$$

$$\text{With } \gamma = \frac{1}{2}(\alpha + \beta) \text{ and } K \text{ the constant terms in the expression.}$$

1.6.3 Potential energy 2dof technical spring mechanism (1.3.2)

Consider the cone-spring mechanism given in Fig.1-28. The mechanism consists of two cones with equal apical angles of 2α . The cones are interconnected with a spherical joint at the apex. On the surfaces of the cones, six lever arms (a to c) are mounted such that their projections on the xy -plane have mutual angles of 120° (neutral position). Three technical linear springs with stiffness C_1, C_2, C_3 and free-lengths l_{01}, l_{02} and l_{03} respectively, are connected to the ends of respectively the lever arms $a \neq b \neq c$. The bottom cone and the coordinate system are considered fixed to the ground.

Figure 1-28: Spring mechanism as shown in section 1.3.2 (Fig.1-21) projected on the xy - and yz -plane. a^* , b^* and c^* are the projections of the levers a , b and c on the yz -plane respectively. a) 3D and top view of the mechanism. b) The mechanism in neutral position. c) The top cone of mechanism rotated through an angle 2θ about the x -axis, relative to the bottom cone.

As was the case for the 2dof balancing mechanism described in section 1.2.2, the potential energy expression of the mechanism will be considered as follows: In stead of rotating the top

cone relative to the bottom cone about an arbitrary axis lying in the xy -plane, to subsequently rotate the top cone relative to the bottom cone through an angle 2θ about that axis, the mechanism -as a whole- is rotated through an angle φ about the z -axis. Subsequently an angle of 2θ about the x -axis is enforced on the top cone and the potential energy (change) of the mechanism is determined.

With rotations 2θ of the top cone, it is seen that two mutual spring attachment points always move in one plane perpendicular to the axis of rotation or, for this consideration, parallel to the yz -plane. Therefore the distance between two mutual spring attachment points can be determined directly from the projection of the mechanism on the yz -plane. Considering linear technical springs are employed, the potential energy of the mechanism then is given by:

$$V = \frac{1}{2}C_1 \cdot (l_1^2 + l_0^2 - 2 \cdot l_0 \cdot l_1) + \frac{1}{2}C_2 \cdot (l_2^2 + l_0^2 - 2 \cdot l_0 \cdot l_2) + \frac{1}{2}C_3 \cdot (l_3^2 + l_0^2 - 2 \cdot l_0 \cdot l_3); \quad (13)$$

With l_1 to l_3 the distance between two mutual spring attachment points.

Figure 1-28 yields the following geometric relations:

$$l_1 = 2 \cdot a^* \cdot \sin(\psi_1); \quad l_2 = 2 \cdot b^* \cdot \sin(\psi_2); \quad l_3 = 2 \cdot c^* \cdot \sin(\psi_3) \quad (14)$$

$$\text{With: } \psi_1 = 0.5\pi - \rho_1 + \theta; \quad \psi_2 = 0.5\pi - \rho_2 - \theta; \quad \psi_3 = 0.5\pi - \rho_3 - \theta$$

The sine terms in equations can be written as:

$$\begin{aligned} \sin(\psi_1) &= \sin(0.5\pi - \rho_1 + \theta) = \cos(\rho_1) \cdot \cos(\theta) + \sin(\rho_1) \cdot \sin(\theta) \\ \sin(\psi_2) &= \sin(0.5\pi - \rho_2 - \theta) = \cos(\rho_2) \cdot \cos(\theta) - \sin(\rho_2) \cdot \sin(\theta); \\ \sin(\psi_3) &= \sin(0.5\pi - \rho_3 - \theta) = \cos(\rho_3) \cdot \cos(\theta) - \sin(\rho_3) \cdot \sin(\theta) \end{aligned} \quad (15)$$

Finally the angles ρ_i can be derived from:

$$\begin{aligned} \sin(\rho_1) &= \frac{a \cdot \sin(\alpha) \cdot \cos(\varphi)}{a^*} & \cos(\rho_1) &= \frac{a \cdot \cos(\alpha)}{a^*} \\ \sin(\rho_2) &= \frac{b \cdot \sin(\alpha) \cdot \cos(\frac{1}{3}\pi - \varphi)}{b^*} & \cos(\rho_2) &= \frac{b \cdot \cos(\alpha)}{b^*} \\ \sin(\rho_3) &= \frac{c \cdot \sin(\alpha) \cdot \cos(\frac{1}{3}\pi + \varphi)}{c^*} & \cos(\rho_3) &= \frac{c \cdot \cos(\alpha)}{c^*} \end{aligned} \quad (16)$$

Substituting eq.15) and eq.16) in eq.14) gives:

$$\begin{aligned} l_1 &= 2 \cdot a \cdot \cos \alpha \cos \theta + 2 \cdot a \cdot \sin \alpha \sin \theta \cos \varphi \\ l_2 &= 2 \cdot b \cdot \cos \alpha \cos \theta - 2 \cdot b \cdot \sin \alpha \sin \theta \cos(\frac{1}{3}\pi - \varphi) \\ l_3 &= 2 \cdot c \cdot \cos \alpha \cos \theta - 2 \cdot c \cdot \sin \alpha \sin \theta \cos(\frac{1}{3}\pi + \varphi) \end{aligned} \quad (17)$$

Thus:

$$\begin{aligned} l_1^2 &= 4 \cdot a^2 \cdot \cos^2 \alpha \cos^2 \theta + 4 \cdot a^2 \cdot \sin^2 \alpha \sin^2 \theta \cos^2 \varphi + 8 \cdot a^2 \cdot \cos \alpha \cos \theta \sin \alpha \sin \theta \cos \varphi \\ l_2^2 &= 4 \cdot b^2 \cdot \cos^2 \alpha \cos^2 \theta + 4 \cdot b^2 \cdot \sin^2 \alpha \sin^2 \theta \cos^2 (\frac{1}{3}\pi - \varphi) - 8 \cdot b^2 \cdot \cos \alpha \cos \theta \sin \alpha \sin \theta \cos (\frac{1}{3}\pi - \varphi) \\ l_3^2 &= 4 \cdot c^2 \cdot \cos^2 \alpha \cos^2 \theta + 4 \cdot c^2 \cdot \sin^2 \alpha \sin^2 \theta \cos^2 (\frac{1}{3}\pi + \varphi) - 8 \cdot c^2 \cdot \cos \alpha \cos \theta \sin \alpha \sin \theta \cos (\frac{1}{3}\pi + \varphi) \end{aligned}$$

With these geometric relations determined, the total potential energy of this mechanism can be determined according to eq.13). Substituting eq.17) into eq.13) gives:

$$\begin{aligned} V &= \frac{1}{2} C_1 \left(4a^2 (\cos^2 \alpha \cos^2 \theta + \sin^2 \alpha \sin^2 \theta \cos^2 \varphi + 2 \cos \alpha \cos \theta \sin \alpha \sin \theta \cos \varphi) \right. \\ &\quad \left. + l_{01}^2 - 4al_{01} (\cos \alpha \cos \theta + \sin \alpha \sin \theta \cos \varphi) \right) \\ &+ \frac{1}{2} C_2 \left(4b^2 (\cos^2 \alpha \cos^2 \theta + \sin^2 \alpha \sin^2 \theta \cos^2 (\frac{1}{3}\pi - \varphi) - 2 \cos \alpha \cos \theta \sin \alpha \sin \theta \cos (\frac{1}{3}\pi - \varphi)) \right. \\ &\quad \left. + l_{02}^2 - 4bl_{02} (\cos \alpha \cos \theta - \sin \alpha \sin \theta \cos (\frac{1}{3}\pi - \varphi)) \right) \\ &+ \frac{1}{2} C_3 \left(4c^2 (\cos^2 \alpha \cos^2 \theta + \sin^2 \alpha \sin^2 \theta \cos^2 (\frac{1}{3}\pi + \varphi) - 2 \cos \alpha \cos \theta \sin \alpha \sin \theta \cos (\frac{1}{3}\pi + \varphi)) \right. \\ &\quad \left. + l_{03}^2 - 4cl_{03} (\cos \alpha \cos \theta - \sin \alpha \sin \theta \cos (\frac{1}{3}\pi + \varphi)) \right) \end{aligned} \quad 18)$$

Re-ordering gives:

$$\begin{aligned} V &= \frac{1}{2} (C_1 \cdot l_{01}^2 + C_2 \cdot l_{02}^2 + C_3 \cdot l_{03}^2) + 2 \cdot \sin^2 \alpha \cdot \sin^2 \theta \cdot \left(C_1 \cdot a^2 \cdot \cos^2 \varphi + C_2 \cdot b^2 \cdot \cos^2 (\frac{1}{3}\pi - \varphi) \right. \\ &\quad \left. + C_3 \cdot c^2 \cdot \cos^2 (\frac{1}{3}\pi + \varphi) \right) \\ &+ 4 \cdot \cos \alpha \cdot \sin \alpha \cdot \cos \theta \cdot \sin \theta \cdot \left(C_1 \cdot a^2 \cdot \cos \varphi - C_2 \cdot b^2 \cdot \cos (\frac{1}{3}\pi - \varphi) \right. \\ &\quad \left. - C_3 \cdot c^2 \cdot \cos (\frac{1}{3}\pi + \varphi) \right) \\ &+ 2 \cdot \cos^2 \alpha \cdot \cos^2 \theta \cdot (C_1 \cdot a^2 + C_2 \cdot b^2 + C_3 \cdot c^2) - 2 \cdot \cos(\alpha) \cdot \cos(\theta) \cdot \left(C_1 \cdot l_{01} \cdot a + C_2 \cdot l_{02} \cdot b \right. \\ &\quad \left. + C_3 \cdot l_{03} \cdot c \right) \\ &- 2 \cdot \sin(\alpha) \cdot \sin(\theta) \cdot \left(C_1 \cdot l_{01} \cdot a \cdot \cos(\varphi) - C_2 \cdot l_{02} \cdot b \cdot \cos(\frac{1}{3}\pi - \varphi) \right. \\ &\quad \left. - C_3 \cdot l_{03} \cdot c \cdot \cos(\frac{1}{3}\pi + \varphi) \right) \end{aligned}$$

It is seen that with $C_1 \cdot a^2 = C_2 \cdot b^2 = C_3 \cdot c^2$ and $C_1 \cdot a \cdot l_{01} = C_2 \cdot b \cdot l_{02} = C_3 \cdot c \cdot l_{03}$ the φ -dependency is eliminated from the equation (condition for rotation symmetric mechanism), and the equation can be simplified to:

$$\begin{aligned} V &= 2 \cdot \sin^2 \alpha \cdot \sin^2 \theta \cdot C_1 \cdot a^2 (\cos^2 \varphi + \cos^2 (\frac{1}{3}\pi - \varphi) + \cos^2 (\frac{1}{3}\pi + \varphi)) \\ &+ 4 \cdot \cos(\alpha) \cdot \sin(\alpha) \cdot \cos(\theta) \cdot \sin(\theta) \cdot C_1 \cdot a^2 (\cos(\varphi) - \cos(\frac{1}{3}\pi - \varphi) - \cos(\frac{1}{3}\pi + \varphi)) \\ &+ 6 \cdot \cos^2 \alpha \cdot \cos^2 \theta \cdot C_1 \cdot a^2 - 6 \cdot C_1 \cdot a \cdot l_{01} \cdot \cos(\alpha) \cdot \cos(\theta) \\ &- 2 \cdot C_1 \cdot a \cdot l_{01} \cdot \sin(\alpha) \cdot \sin(\theta) \cdot (\cos(\varphi) - \cos(\frac{1}{3}\pi - \varphi) - \cos(\frac{1}{3}\pi + \varphi)) + K \end{aligned} \quad 19)$$

With K the constant terms in the equation

Since:

$$\begin{aligned} \cos^2 (\frac{1}{3}\pi - \varphi) + \cos^2 (\frac{1}{3}\pi + \varphi) &= \frac{1}{2} \cos^2 \varphi + 1.5 \sin^2 \varphi \\ \cos(\frac{1}{3}\pi - \varphi) + \cos(\frac{1}{3}\pi + \varphi) &= \cos \varphi \end{aligned}$$

Equation 19) can be re-written to:

$$\begin{aligned} V &= 3 \cdot \sin^2 \alpha^2 \cdot \sin^2 \theta^2 \cdot C_1 \cdot a^2 + 6 \cdot \cos^2 \alpha \cdot \cos^2 \theta \cdot C_1 \cdot a^2 - 6 \cdot C_1 \cdot a \cdot l_{01} \cdot \cos \alpha \cdot \cos \theta + K \Rightarrow \\ V &= 3C_1 a^2 \left(\frac{1}{2} \cos^2 \alpha (2 - \tan^2 \alpha) \cos(2\theta) + 1 - 0.5 \sin^2 \alpha \right) - 6C_1 a l_{01} \cdot \cos \alpha \cdot \cos \theta + K \Rightarrow \\ V &= 1.5C_1 a^2 \cos^2 \alpha (2 - \tan^2 \alpha) \cos(2\theta) - 6C_1 a l_{01} \cdot \cos \alpha \cdot \cos \theta + K \end{aligned} \quad 20)$$

With K the constant terms; $C_1 \cdot a^2 = C_2 \cdot b^2 = C_3 \cdot c^2$; and $C_1 \cdot a \cdot l_{01} = C_2 \cdot b \cdot l_{02} = C_3 \cdot c \cdot l_{03}$

Equation 20 is the potential energy expression of the three-spring mechanism shown in Fig.1-28 for rotations 2θ of the top cone relative to the bottom cone about an arbitrary axis lying in the xy -plane.

1.6.4 Torsion stability 2dof balancing mechanisms

This section shortly discusses the torsion stability or torsion stiffness of some of the mechanisms considered in this chapter. This is done by means of potential energy considerations.

Mechanism shown in Fig.1-21 and Fig.1-22

Considering the gravity balancing mechanism given in Fig.1-21 and 1-29, it is seen that the top cone rotates about two axis going through the apexes of the cones, parallel to the xy -plane (also see Fig.1-30). The whole top cone –inclusive sphere- is rotated through an angle θ about an axis going through the apex of the bottom cone, but the top cone itself is simultaneously rotated through an angle θ about an axis going through its' own apex. Having determined the static balancing conditions for these rotations in the previous section, another subsequent possible motion of the top cone relative to the bottom cone -assuming only rolling of the spheres- is a rotation ψ about the connection line between the centres of the spheres. This means it is important to verify whether or not the assumption that a pure rotation of the top cone about these two parallel axis, parallel to the xy -plane is possible without causing torsion instability about the connection line of the cones' centres. This will be done by means of a potential energy consideration.

Using vector notation, the positions of the spring attachment points on the bottom cone are described by a global coordinate system $[x, y, z]$, which is considered fixed to the ground and to the apex of the bottom cone (Fig.1-29). The paths of the spring attachment points on the top cone are defined relative to a coordinate system $[x_{tc}, y_{tc}, z_{tc}]$, which has the same orientation as the global coordinate system but is considered fixed to the apex of the top cone.

Figure 1-29: a) and b) 3D view of mechanism shown in Fig.1-21 with the mechanism in neutral position and rotated –as a whole- through an angle φ about the z -axis. c) Top view of mechanism.

With the mechanism in the neutral position and rotated –as a whole- through a angle φ about the z -axis, it is seen that the lever-arm vectors relative to the coordinate systems are given by (Fig.1-29):

$$\bar{v}_a = \begin{bmatrix} x_{tc} \\ y_{tc} \\ z_{tc} \end{bmatrix} = \begin{bmatrix} -a \cdot \sin \alpha \cdot \sin \varphi \\ -a \cdot \sin \alpha \cdot \cos \varphi \\ a \cdot \cos \alpha \end{bmatrix}; \quad \bar{v}_e = \begin{bmatrix} b \cdot \sin \alpha \cdot \sin(\frac{1}{3}\pi + \varphi) \\ b \cdot \sin \alpha \cdot \cos(\frac{1}{3}\pi + \varphi) \\ b \cdot \cos \alpha \end{bmatrix}; \quad \bar{v}_c = \begin{bmatrix} -c \cdot \sin \alpha \cdot \sin(\frac{1}{3}\pi - \varphi) \\ c \cdot \sin \alpha \cdot \cos(\frac{1}{3}\pi - \varphi) \\ c \cdot \cos \alpha \end{bmatrix}; \quad \bar{R}_1 = \begin{bmatrix} 0 \\ 0 \\ R_1 \end{bmatrix} \quad 21)$$

and:

$$\bar{v}_b = \begin{bmatrix} x \\ y \\ z \end{bmatrix} = \begin{bmatrix} -a \cdot \sin \beta \cdot \sin \varphi \\ -a \cdot \sin \beta \cdot \cos \varphi \\ -a \cdot \cos \beta \end{bmatrix}; \quad \bar{v}_d = \begin{bmatrix} -b \cdot \sin \beta \cdot \sin(\frac{1}{3}\pi - \varphi) \\ b \cdot \sin \beta \cdot \cos(\frac{1}{3}\pi - \varphi) \\ -b \cdot \cos \beta \end{bmatrix}; \quad \bar{v}_f = \begin{bmatrix} c \cdot \sin \beta \cdot \sin(\frac{1}{3}\pi + \varphi) \\ c \cdot \sin \beta \cdot \cos(\frac{1}{3}\pi + \varphi) \\ -c \cdot \cos \beta \end{bmatrix}$$

Figure 1-30: Mechanism as shown in Fig.1-29. a) Mechanism in neutral position projected on the yz -plane. b) Top cone rotated through an angle 2θ about its x_{ic} -axis. c) As b) but with the top cone rotated through an angle ψ about the connection-line of the centres of the spheres (=apexes cones).

The position of the vectors on the top cone relative to the $[x_{ic}, y_{ic}, z_{ic}]$ -coordinate system after the subsequent rotations 2θ about the x -axis followed by the rotation ψ about the z'_{ic} axis that is rotated through an angle θ about the x_{ic} axis (extension of the connection line of the apexes of the two cones) is described by the following matrix operation (Fig.1-30):

$$\bar{v}' = \begin{bmatrix} x'_{ic} \\ y'_{ic} \\ z'_{ic} \end{bmatrix} = [R_x] \bullet [R_{z'_{ic}}] \bullet [R_x] \bullet \bar{v} = [R_x] \bullet [R_{z'_{ic}}] \bullet [R_x] \bullet \begin{bmatrix} x_{ic} \\ y_{ic} \\ z_{ic} \end{bmatrix} \quad (22)$$

With: $[R_{z'_{ic}}]$ and $[R_x]$:

$$[R_{z'_{ic}}] = \begin{bmatrix} \cos \psi & \sin \psi & 0 \\ -\sin \psi & \cos \psi & 0 \\ 0 & 0 & 1 \end{bmatrix} \text{ and } [R_x] = \begin{bmatrix} 1 & 0 & 0 \\ 0 & \cos \theta & \sin \theta \\ 0 & -\sin \theta & \cos \theta \end{bmatrix} \quad (23)$$

The vectors that span two mutual spring attachment points and the vector that goes from the apex of the bottom cone to the mass can now be determined according to:

$$\begin{aligned}
 \bar{\mathbf{v}}_a - \bar{\mathbf{v}}_b + \bar{\mathbf{v}}_{shift} &= \bar{\mathbf{v}}_{length_1} \\
 \bar{\mathbf{v}}_c - \bar{\mathbf{v}}_d + \bar{\mathbf{v}}_{shift} &= \bar{\mathbf{v}}_{length_2} \\
 \bar{\mathbf{v}}_e - \bar{\mathbf{v}}_f + \bar{\mathbf{v}}_{shift} &= \bar{\mathbf{v}}_{length_3} \\
 \bar{\mathbf{R}}_1 + \bar{\mathbf{v}}_{shift} &= \bar{\mathbf{v}}_{mass}
 \end{aligned} \tag{24}$$

With $\mathbf{v}_{shift}$ the translation vector from $[x, y, z]$ to $[x_{tc}, y_{tc}, z_{tc}]$:

$$\bar{\mathbf{v}}_{shift} = \begin{bmatrix} 0 \\ R_2 \sin \theta \\ R_2 \cos \theta \end{bmatrix}$$

With all vectors determined, the total potential energy of the mechanism can be determined according to:

$$\begin{aligned}
 V_{tot} &= \frac{1}{2} C1 \cdot \left(\sqrt{\bar{\mathbf{v}}_{length_1}^T \cdot \bar{\mathbf{v}}_{length_1}} - l_{01} - R_2 \right)^2 + \frac{1}{2} C2 \cdot \left(\sqrt{\bar{\mathbf{v}}_{length_2}^T \cdot \bar{\mathbf{v}}_{length_2}} - l_{02} - R_2 \right)^2 \\
 &+ \frac{1}{2} C3 \cdot \left(\sqrt{\bar{\mathbf{v}}_{length_3}^T \cdot \bar{\mathbf{v}}_{length_3}} - l_{03} - R_2 \right)^2 + Pot_m
 \end{aligned} \tag{25}$$

With Pot_m the contribution of the mass to the potential energy of the system:

$$Pot_m = \bar{\mathbf{v}}_{mass}^T \cdot \begin{bmatrix} 0 \\ 0 \\ mg \end{bmatrix}$$

To determine whether or not the mechanism is torsion stable, two derivatives of the total potential energy expression to the rotation angle ψ are taken. The first should be equal to zero for $\psi=0$ and the second derivative should be greater then zero for $\psi=0$. Thus:

$$\begin{aligned}
 \frac{\partial V_{tot}}{\partial \psi} &= 0 \\
 \frac{\partial^2 V_{tot}}{\partial^2 \psi} &> 0
 \end{aligned}$$

Substitution of the measures of the prototype shown in Fig.1-22 in eq.25) and taking the derivative to the rotation ψ of the top cone relative to the bottom cone (about the extension of the connection line of the apexes of the cones) twice, will show that the conditions set out above are satisfied; at least for the range of motion of the mechanism (maximum rotation θ is where the tension springs have zero tension). Thus the mechanism is torsion stable, which is also seen in practice.

Mechanism shown in Fig.1-24

Since the mechanism shown in Fig.1-24 and Fig.1-28 is basically the same as the mechanism described in section 1.2.2, the same method can be used to determine the torsion stability of these mechanisms. The only difference is that technical springs are employed instead of ideal springs, meaning that the strain-length of the springs changes, and the potential energy of the mechanism changes to:

$$V_{tot} = \frac{1}{2}C_1 \cdot \left(\sqrt{\bar{\mathbf{v}}_{length_1}^T \bullet \bar{\mathbf{v}}_{length_1}} - l_{01} \right)^2 + \frac{1}{2}C_2 \cdot \left(\sqrt{\bar{\mathbf{v}}_{length_2}^T \bullet \bar{\mathbf{v}}_{length_2}} - l_{02} \right)^2 + \frac{1}{2}C_3 \cdot \left(\sqrt{\bar{\mathbf{v}}_{length_3}^T \bullet \bar{\mathbf{v}}_{length_3}} - l_{03} \right)^2 \quad 26)$$

Of course the mass, being mounted on the axis of revolution, has no influence on the potential energy and therefore does not need to be considered.

To determine whether or not the mechanism is torsion stable, two derivatives of the total potential energy expression to the rotation angle ψ are taken. The first should be equal to zero for $\psi=0$ and the second derivative should be greater than zero for $\psi=0$. Thus:

$$\frac{\partial V_{tot}}{\partial \psi} = 0$$

$$\frac{\partial^2 V_{tot}}{\partial^2 \psi} > 0$$

Substitution of the measures of the prototypes shown in Fig.1-24 in eq.26) and taking the derivative to the rotation ψ of the top cone (about the axis of revolution of the top cone) twice, learns that the conditions set out above are satisfied; at least for the range of motion of the mechanism (maximum rotation θ is where the tension springs have zero tension). Thus the mechanism is torsion stable, which is of -again- also seen in practice.