Transverse tripolar spinal cord stimulation: theoretical performance of a dual channel system

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Abstract—*A new approach to spinal cord stimulation is presented, by which several serious problems of conventional methods can be solved. A transverse tripolar electrode with a dual-channel voltage stimulator is evaluated theoretically by means of a volume conductor model, combined with nerve fibre models. The simulations predict that a high degree of freedom in the control of activation of dorsal spinal pathways may be obtained with the described system. This implies an easier control of paraesthesia coverage of skin areas and the possibility to correct undesired paraesthesia patterns, caused by lead migration, tissue growth, or anatomical asymmetries, for example, without surgical intervention. It will also be possible to preferentially activate either dorsal column or dorsal root fibres, which has some important clinical advantages. Compared to conventional stimulation systems, the new system has a relatively high current drain.*

Keywords—*Electrical stimulation, Electrode, Field steering, Finite difference model, Nerve fibre model, Paraesthesia, Potential field, Spinal cord stimulation*

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1 Introduction

IN SPINAL cord stimulation (SCS), an electrode is usually used, consisting of a four-contact array, oriented rostrocaudally in the dorsal epidural space. If each combination has at least one cathode, the four-contact array provides 65 possible anode–cathode combinations. Together with freely accessible parameters such as pulse amplitude, duration and rate, and the medio-lateral and rostrocaudal positions of the electrode, there is some flexibility in directing paraesthesia coverage and intensity.

However, major problems exist, which are related to the positioning and stabilisation of the paraesthesiae. With current implantable SCS systems, it is difficult to place the lead in such a way that optimal paraesthesia coverage is obtained, which is a serious drawback because of the widely recognised significance of a complete coverage of the painful area in SCS for pain management (SHEALY *et al.*, 1970; SWEET and WEPSIG, 1974; NIELSON *et al.*, 1975; BURTON, 1977; KRAINICK *et al.*, 1980; MAIMAN *et al.*, 1985; LAW, 1986; VOGEL *et al.*, 1986; BAROLAT *et al.*, 1991; NORTH *et al.*, 1991). Problems, such as lead migration, change of the pain topography, low discomfort thresholds relative to the perception thresholds, muscle contractions and histological changes, such as the growth of connective tissue around the electrode, also add to the limitations of SCS (SHARKEY, 1981; LAW, 1983; 1986; BAROLAT *et al.*, 1991). Thus, even if during surgery paraesthesiae cover the pain area totally, the paraesthesia pattern often changes afterwards. Therefore, it would be highly desirable to be able to refocus paraesthesiae after surgery.

Two approaches have been adopted to reduce the paraesthesia coverage problem. A relatively large flat electrode, which requires a laminotomy for implantation, can be positioned accurately during surgery and, because of its shape, the electrode hardly migrates (BAROLAT *et al.*, 1991). The second approach is the use of multiple cylindrical electrode arrays in parallel, to increase the number of possible contact combinations, thus improving the probability of obtaining proper paraesthesia coverage and correcting lead migration (LAW, 1986).

A multi-electrode approach may also be needed to reduce the occurrence of motor responses or other uncomfortable segmentary effects. Both muscle contractions and other segmentary effects are probably caused by direct stimulation of dorsal roots (radicular stimulation) (BANTLI *et al.*, 1975; DIMITRIJEVIC *et al.*, 1980; COBURN, 1985; STRUIJK *et al.*, 1993a). If after surgery the paraesthesia pattern is perfect, the stimulation may still be useless because of the occurrence of motor responses at voltages below the therapeutically effective values (BANTLI *et al.*, 1975; BAROLAT *et al.*, 1991). This occurs mainly at mid-thoracic levels, causing a so-called segmentary band (BAROLAT *et al.*, 1991), but sometimes at other levels as well.

The problems of paraesthesia coverage and motor responses may be overcome by a stimulation method, based on a transverse tripolar contact configuration with two independently controlled voltage stimulation channels. In this work we

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illustrate a tripolar stimulation method and its performance with the help of a computer model of the spinal cord and the most relevant neural elements. We show that the proposed system creates the possibility of changing the paraesthesia pattern from symmetrical to asymmetrical, and *vice versa*, and also of correcting the paraesthesia pattern in cases of lead migration, asymmetrical spinal cord position and anatomical changes. We also show that the activation of dorsal root fibres can be reduced in favour of dorsal column fibres, thus presumably reducing the occurrence of radicular effects.

2 Methods

2.1 *Dual-channel transverse tripolar system**

Fig. 1 is a schematic drawing of the transverse tripolar lead with its connections to the pulse generators, which are voltage sources in all commercially available stimulators because of practical advantages over current sources (see Section 4). The dimensions used in the modelling study are shown in the Figure. The lead consists of three metal strips (contacts). The central contact (referred to as the cathode, although it can also be used as an anode) is somewhat shorter than the two lateral contacts (anodes). Two voltage sources V_1 and V_2 are connected to the lead. V_1 is connected to one anode and the cathode. V_2 is connected to the second anode and the cathode.

The lengths and widths of the three contacts and the contact spacings are the main lead parameters. In this study these dimensions are fixed. The following design criteria were applied.

2.1.1 *Contact area:* this should be at least 12 mm^2, because such a contact area has been approved in commercially available leads; according to stimulation amplitudes in current SCS practice (amplitude < 10 V with a typical 1 kΩ load), average current densities will thus be less than 1 mA mm^{-2} at the electrode surface.

2.1.2 *Contact length:* anodal contacts should be longer than the central cathode to maintain a good shielding effect, even if the lead is somewhat rotated in the coronal plane, which may occur if the electrode has moved or if it was not implanted perfectly rostrocaudally.

2.1.3 *Contact spacing:* this should be larger than the thickness

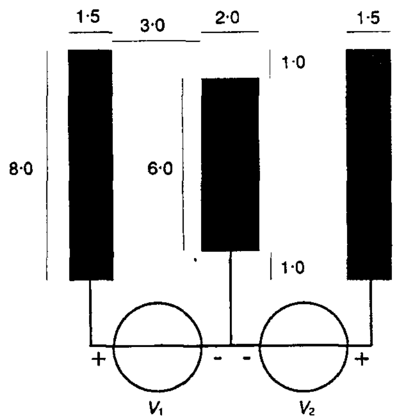

Fig. 1 *Transverse tripolar lead with two voltage sources; dimensions are in mm*

* US Patent number: 5,501,703

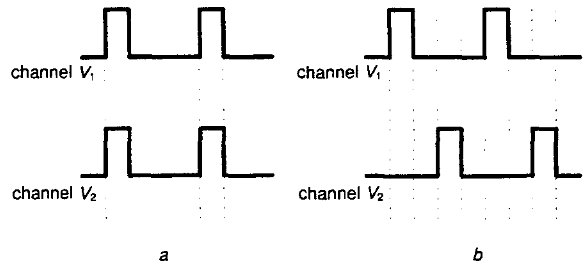

Fig. 2 *Timing of the two channels in (a) simultaneous operating mode; (b) alternating operating mode*

of the dorsal cerebrospinal fluid (CSF) layer; otherwise high stimulus amplitudes will be required due to the shunting effect of the highly conductive CSF (HOLSHEIMER and STRUIJK 1991, HOLSHEIMER *et al.*, 1995). It should be approximately the distance between the dorsal root entry zone and the spinal cord mid-line. The thickness of the CSF layer and the spinal cord geometry thus affect the optimal contact spacing for different spinal levels (HOLSHEIMER *et al.*, 1994).

2.1.4 *Total lead width:* this should be as small as possible to minimise the surgical procedure.

2.1.5 *Contact width:* as a consequence of the criteria for contact spacing and total lead width, contact width should be kept small.

The voltage sources are pulse generators with independent amplitude control. The system can be used in two operating modes.

(i) Simultaneous mode (Fig. 2*a*): in this mode, the system is essentially tripolar, and this mode is mainly used in this work.
(ii) Alternating mode: the pulses of one channel are interleaved with those of the other (Fig. 2*b*). In this mode, the system acts as two independent bipolar systems.

In monopolar stimulation only the central contact of the lead is used. The second contact is the metal can of the implantable pulse generator, which is modelled as a distant ground in the volume conductor model (see Section 2.2).

2.2 *Computer models*

The spinal cord stimulation model consists of two parts. The first part is a 3-D conductor model of the spinal cord and its surroundings. This model comprises the major macro-anatomical structures and the stimulating electrodes. The second part consists of models of large myelinated dorsal root and dorsal column nerve fibres. The dorsal roots and dorsal columns are the spinal cord structures closest to the electrodes, which are placed in the dorsal epidural space, and will therefore be the primary targets of the stimulation. These models have been described previously (HOLSHEIMER *et al.*, 1991; HOLSHEIMER and STRUIJK, 1991; STRUIJK *et al.*, 1991; 1992; 1993a; 1993b).

To assess the direct effects of stimulation on the nerve fibres, a two-step procedure is followed. First, the potential field in the volume conductor model is calculated. Secondly, this field is applied to the nerve fibre models to determine which fibres are excited by the stimulation.

A three-dimensional conductor model of the mid-cervical spinal cord (C4–C6) is used as the basic model. In one simulation, a model of the mid-thoracic spinal cord (T5–T6) is used. Fig. 3 shows transverse sections of these models. The models comprise the spinal cord, which is composed of grey matter (GM) and white matter (WM), cerebrospinal fluid (CSF), epidural space (ES), vertebral bone (VB), a layer representing surrounding tissues (SL), contact insulation (IS) and a thin layer representing the dura mater (DM) at the dorsal side. The small dorsal root filaments, immersed in the highly conductive CSF, are not incorporated in the volume conductor model. The model and the conductivities in the model have been described previously (STRUIJK *et al.*, 1993b).

The electrode contacts in the model are given constant voltages. In monopolar stimulation the boundary of the model serves as the distant anode. The thickness of the dorsal CSF layer was measured from MR images obtained from 26 normal subjects (HOLSHEIMER *et al.*, 1994). In the mid-cervical and the mid-thoracic models these thicknesses are 2.3 mm and 5.8 mm, respectively.

The volume conductor is discretised using a rectangular grid with variable grid spacing. The number of grid points is $57 \times 57 \times 57 = 185193$. A finite difference method is applied to discretise the governing Laplace equation. The resulting set of linear equations is solved using a Red–Black Gauss–Seidel iteration with variable over-relaxation.

Two types of nerve fibres were modelled: the dorsal root (DR) fibre and dorsal column (DC) fibre. DC fibres are longitudinal fibres in the dorsal columns that issue collaterals into the grey matter.

The model of a DC fibre is a cable model (MCNEAL, 1976) extended with collaterals. Close to the electrode, collaterals are attached to each second node of Ranvier of the modelled 21-noded fibre (STRUIJK *et al.*, 1992). For DR fibres a cable model with a curved trajectory is used, with the proximal end connected to a DC fibre model (STRUIJK *et al.*, 1993a).

Table 1 shows the geometric fibre parameters; the electrical parameters have been established previously (STRUIJK *et al.*, 1992). In all simulations a monophasic rectangular pulse with a duration of 210 μs was used.

2.3 *Evaluation parameters*

To compare the effects of different stimulus parameters in various settings, several evaluation parameters were defined. In these parameter definitions, the threshold stimulus is defined as the lowest stimulus amplitude at which the (10 μm diameter) fibre under concern is excited when a pulse of 210 μs duration is used.

Apart from these parameters, plots of the recruited areas for DC fibres show how asymmetrical or how selective the stimulation is. In a transverse section the recruited area is bordered by the spinal cord's dorsal boundary and by a recruitment contour line more ventrally in the dorsal columns. The recruited area is then the area in which 10 μm DC fibres are excited at a given stimulus amplitude. In general, recruitment contour lines do not have the same shape as isopotential lines, and thus the isopotential lines do not always reflect the delineation of the recruited area.

Table 1 Fibre parameters

	DC fibre	DR fibre
diameter of main fibre	10 μm	10 μm
diameter of collaterals	3·33 μm	—
inner fibre diameter	0·6 × fibre diameter	0·6 × fibre diameter
length of node of Ranvier	1·5 μm	1·5 μm
number of nodes of main fibre	21	21
number of nodes per collateral	8	—
number of collaterals	7	—
internodal length	100 × fibre diameter	100 × fibre diameter
collateral spacing	2 × internodal length	—

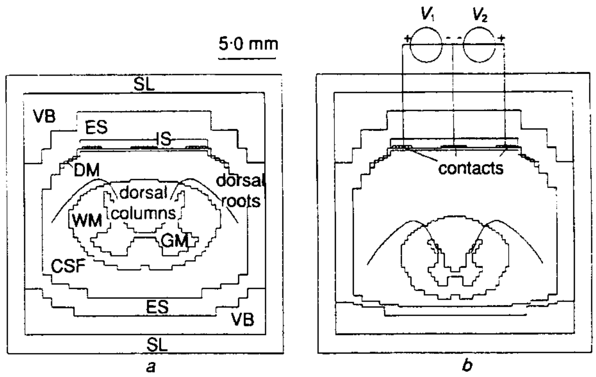

Fig. 3 *Transverse section of (a) mid-cervical spinal cord model; (b) mid-thoracic model; the connections of the contacts to the voltage sources are drawn schematically*

The following evaluation parameters were defined.

V_{DC} = threshold stimulus of a DC fibre at the dorsomedial boundary of the spinal cord.

V_{DR} = threshold stimulus of a DR fibre entering the cord at the rostrocaudal level corresponding to the centre of the cathode; if the simulation is asymmetrical, then the fibre with the lowest threshold (either the left or right DR fibre) is taken.

Pr = fibre type preference; Pr is defined as $Pr = V_{DC}/V_{DR}$; therefore, $Pr > 1$ ($Pr < 1$) means that DR fibres (DC fibres) are more likely to be excited.

As_{DC} = left–right asymmetry coefficient for dorsal column fibres; $As_{DC} = V_{DC,left}/V_{DC,right}$. $V_{DC,left}$ ($V_{DC,\ right}$) is the threshold of a DC fibre at the dorsal border of the dorsal columns, at 2.5 mm left (right) of mid-line. If $As_{DC} > 1$ ($As_{DC} < 1$), the stimulation is asymmetrical and thresholds are lower at the right (left) side.

As_{DR} = left-right asymmetry coefficient for dorsal root fibres; $As_{DR} = V_{DR,left}/V_{DR,right}$. $V_{DR,left}$ ($V_{DR,right}$) is the threshold of the left (right) DR fibre. If $As_{DR} > 1$ ($As_{DR} < 1$), then the stimulation is asymmetrical and thresholds are lower at the right (left) side.

W = the width of the recruited dorsal column area. To calculate W we choose the stimulus amplitude such that a DC fibre at midline and 1·0 mm from the dorsal boundary in the dorsal columns is just excited. This amplitude is then used to obtain the recruitment contour. W is defined as the distance between the two intersection points of the recruitment contour line and the dorsal boundary of the spinal cord.

I (mA) = stimulus current needed to excite the dorsomedial DC fibre.

3 Results

3.1 *Monopolar compared with tripolar stimulation*

The marked difference between the potential fields of a monopolar and a tripolar configuration (simultaneous mode, symmetrical stimulation) is illustrated for the mid-cervical model in Figs. 4*a* and *b*. These Figures show isopotential lines in a transverse section at the level of the centre of the contacts. The monopolar and tripolar electrodes are positioned symmetrically with respect to the spinal cord mid-line.

Figs. 4*c* and *d* show the recruited areas ($D = 1{\cdot}0$ mm). Figs. 4*a–d* show that, with tripolar stimulation, the potential field and the recruited area are more restricted to the medial part of the dorsal columns than with monopolar stimulation.

Figs. 4*e* and *f* show the recruited areas in monopolar and tripolar stimulation of the mid-thoracic model. The main difference between the mid-cervical and the mid-thoracic

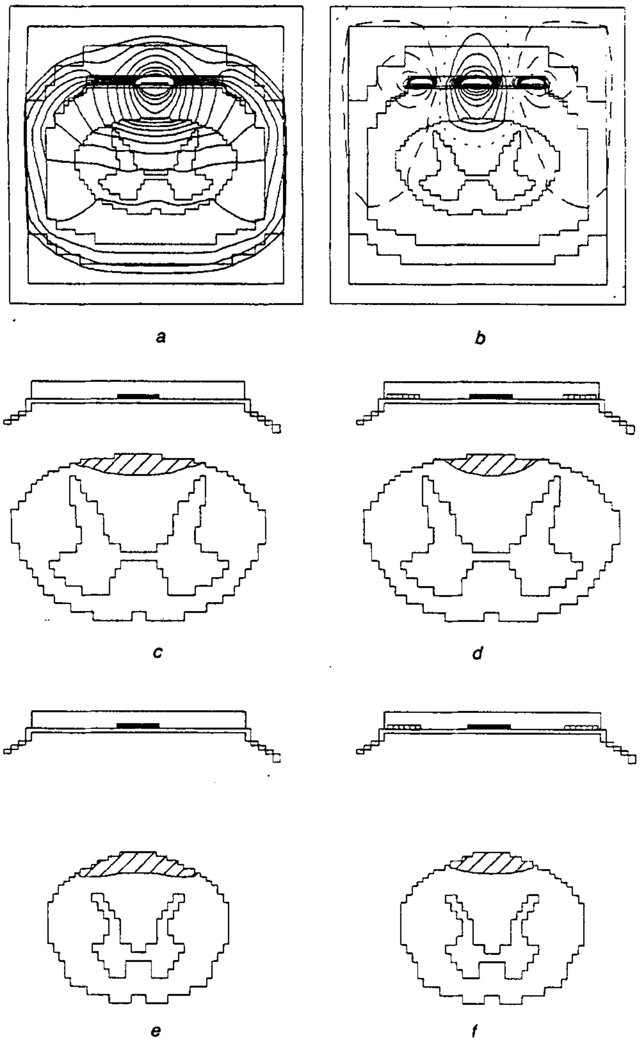

Fig. 4 *Potential fields in the mid-cervical model; (a) monopolar; (b) transverse tripolar stimulation; solid lines = negative; dashed lines = positive; dotted line = zero potential; recruited areas = shaded; (c) mid-cervical monopolar; (d) mid-cervical transverse tripolar; (e) mid-thoracic monopolar; (f) mid-thoracic transverse tripolar stimulation*

models is the difference in the thickness of the dorsal CSF layer, and therefore the difference in distance between contacts and the spinal cord, resulting in higher threshold stimuli in the mid-thoracic model (HOLSHEIMER *et al.*, 1991; STRUIJK *et al.*, 1993b).

A quantitative comparison of monopolar and transverse tripolar stimulation is given in Table 2. From this Table the following conclusions may be drawn.

In tripolar stimulation, both V_{DC} and V_{DR} are higher than in monopolar stimulation. In particular, the tripolar V_{DR} is high. This is reflected in the parameter *Pr*, which shows a preference

Table 2 Evaluation parameters in monopolar and symmetrical transverse tripolar stimulation

	mid-cervical model		mid-thoracic model	
	monopolar	tripolar	monopolar	tripolar
V_{DC}	2·01	$V_1 = V_2 = 3{\cdot}36$	7·86	$V_1 = V_2 = 22{\cdot}9$
V_{DR}	1·47	$V_1 = V_2 = 5{\cdot}43$	3·80	$V_1 = V_2 = 20{\cdot}3$
Pr	1·37	0·62	2·07	1·13
W, mm	7·14	5·0	5·88	4·0
I, mA	3·41	4·97	13·7	33·9

for DR fibres in the monopolar case and, in contrast, a preference for DC fibres in the tripolar configuration. Consequently, in transverse tripolar stimulation it is less likely that motor reflex loops will be activated because the stimulation is more confined to the dorsal columns. In both monopolar and tripolar stimulation, DR fibre preference is higher in the mid-thoracic model than in the mid-cervical model.

W reflects the phenomenon that in transverse tripolar stimulation the width of the recruited area is smaller than in monopolar stimulation. This is also shown in Fig. 4. Therefore, DC fibre thresholds in the lateral parts of the dorsal columns are increased, as compared with thresholds in the medial parts. For the same reason, it is unlikely that fibres in the dorsolateral columns will be activated in transverse tripolar stimulation.

In mid-cervical tripolar stimulation, the current drain *I* is about 40% higher than in monopolar stimulation. In the mid-thoracic model the increase is about 150%.

If the monopole is replaced by any rostrocaudal (longitudinal) multipolar electrode configuration, the transverse shape of the field and the value of *W* would hardly change (HOLSHEIMER *et al.*, 1991).

3.2 *Asymmetrical sources*

In the preceding Section both the volume conductor model and the two voltage sources were symmetrical, and consequently the potential fields and the recruited areas were symmetrical as well ($As_{DR} = As_{DC} = 1{\cdot}00$). With a multipolar longitudinal configuration it is impossible to change a symmetrical into an asymmetrical stimulation in a controllable way, without changing the lead's position.

As shown in Fig. 5, asymmetrical stimulation is easily obtained using unbalanced sources in the transverse tripolar configuration, even if the electrode position is perfectly symmetrical. In this simulation $V_2 = 0{\cdot}0$ V, where V_2 is the voltage between the anode on the right and the central cathode (Fig. 3*b*).

Table 3 (left columns) shows the evaluation parameters of the transverse tripolar lead with balanced sources and with unbalanced sources. The Table shows a marked drop in V_{DR} in the case of unbalanced sources and thus a high preference for DR fibres. The asymmetry coefficients also change dramatically, especially for the DC fibres.

If the lead is not at the mid-line (due to lead migration or to lateral positioning during surgery), it is still possible to obtain an almost symmetrical stimulation. Fig. 6 shows the potential fields and the recruited areas, where the centre of the lead is 1·0 mm from the mid-line. In Figs. 6*a* and *b* the voltage sources have equal amplitudes ($V_1 = V_2$) and the stimulation is clearly asymmetrical.

Figs. 6*c* and *d* show the same geometrical configuration with unbalanced sources: $V_2 = 2V_1$. Although the potential field in

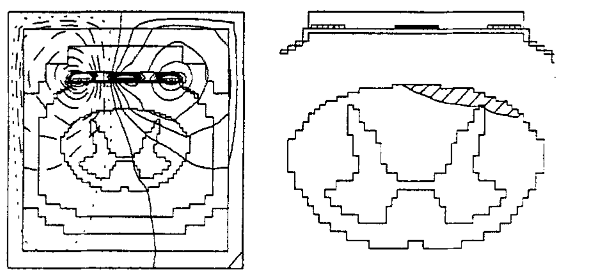

Fig. 5 *Symmetrical electrode position and unbalanced sources ($V_2 = 0$) in the mid-cervical model: (a) potential field; (b) recruited area*

Table 3 Transverse tripole with balanced and unbalanced sources

	symmetrical electrode position		asymmetrical electrode position	
	balanced sources	unbalanced sources	balanced sources	unbalanced sources
V_{DC}	$V_1 = V_2 = 3{\cdot}36$	$V_1 = 6{\cdot}73$; $V_2 = 0{\cdot}0$	$V_1 = V_2 = 3{\cdot}85$	$V_1 = 2{\cdot}33$; $V_2 = 4{\cdot}65$
V_{DR}	$V_1 = V_2 = 5{\cdot}43$	$V_1 = 2{\cdot}51$; $V_2 = 0{\cdot}0$	$V_1 = V_2 = 2{\cdot}92$	$V_1 = 2{\cdot}26$; $V_2 = 4{\cdot}52$
Pr	0·62	2·68	1·3	1·03
As_{DC}	1·00	0·11	0·20	1·02
As_{DR}	1·00	0·57	0·36	0·92

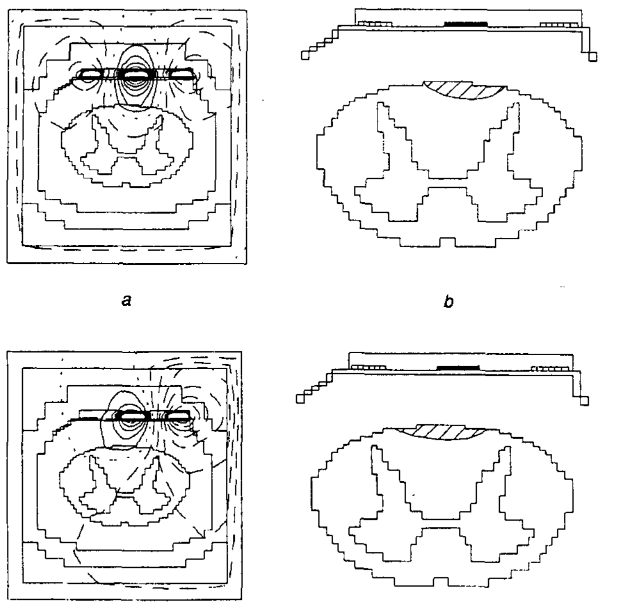

Fig. 6 *Transverse tripolar lead 1·0 mm laterally in the mid-cervical model; (a) potential field; (b) recruited area with balanced sources ($V_1 = V_2$); (c) potential field, (d) recruited area with unbalanced sources ($V_2 = 2V_1$)*

Fig. 6*c* is much more asymmetrical than in Fig. 6*a*, the recruited area in Fig. 6*d* is almost symmetrical.

The results in Table 3 (right columns) show that symmetry is almost completely restored (asymmetry coefficients As_{DC} and As_{DR} are close to 1), whereas fibre type preference Pr is only slightly affected.

3.3 *Asymmetrical spinal cord*

If the spinal cord and not the lead is asymmetrically positioned in the spinal canal, the stimulation will also be asymmetrical. An asymmetrical spinal cord position of 1–2 mm will occur in at least 40% of the patients (HOLSHEIMER *et al.*, 1994). Clinically, this may cause unexpected asymmetrical paraesthesiae if the lead is at the radiological mid-line (BAROLAT *et al.*, 1991). The modelling results are similar to those in the case of a lateral lead, because the geometrical relation between electrode and spinal cord is the same in both cases.

3.4 *Asymmetrical load*

After surgery connective tissue will grow around the lead and other histological changes may occur, possibly yielding an asymmetrical electrical load of the contacts. An asymmetrical load is modelled by decreasing by a factor 2 the conductivity of the dura mater in contact with the right anode (0·015 instead of 0·030 S m^{-1}).

Table 4 Transverse tripole with balanced and unbalanced sources and an asymmetrical load

	balanced sources	unbalanced sources
V_{DC}, V	$V_1 = V_2 = 3{\cdot}37$	$V_1 = 3{\cdot}06$; $V_2 = 4{\cdot}08$
V_{DR}, V	$V_1 = V_2 = 3{\cdot}98$	$V_1 = 4{\cdot}78$; $V_2 = 6{\cdot}37$
Pr	0·85	0·63
As_{DC}	0·72	0·92
As_{DR}	0·77	0·97

A slightly asymmetrical recruited area is obtained when the voltage sources are balanced. Restoration of symmetry is realised by unbalancing the voltage sources ($V_1 = 0{\cdot}75\ V_2$), resulting in a symmetrical recruited area.

The evaluation parameters in Table 4 show that both the asymmetry coefficients and the fibre type preference Pr with unbalanced sources are almost the same as in the symmetrical balanced situation (Table 3).

3.5 *Simultaneous compared with alternating mode*

So far we have illustrated only the possibilities of the tripolar configuration in the simultaneous operating mode. However, the alternating mode may also be useful, especially when larger parts of the body should be covered with paraesthesiae, but only if no problems occur due to motor responses.

Fig. 7 shows the recruited areas obtained with the simultaneous and the alternating operating modes (Fig. 7*a* is identical to Fig. 4*d*). The recruited area in the simultaneous mode is clearly smaller than that realised with the alternating mode. In the simultaneous mode the tripolar potential field is the result of a superposition of two bipolar fields (both bipolar fields are present at the same time). The resulting potential field is narrow and so is the recruited area. This superposition of the potential field occurs only if both bipolar fields are present at the same time, as happens with the stimulus sequence of Fig. 2*a*, but not in the alternating mode (Fig. 2*b*). In the alternating mode the recruited area consists of the union of two (overlapping) recruited areas, as shown in Fig. 7*b*. Fibres in area 1 are recruited when channel V_1 (Fig. 3*b*) is active,

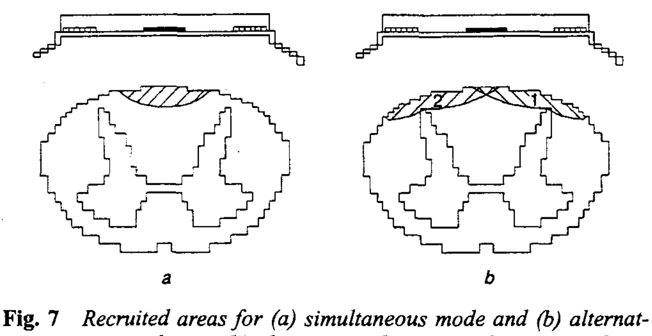

Fig. 7 *Recruited areas for (a) simultaneous mode and (b) alternating mode; in (b) the recruited area is the union of two different areas*

whereas fibres in area 2 are excited when channel V_2 is active. This implies that the fibres in the overlap of areas 1 and 2 are activated at twice the frequency of each channel.

In both operating modes the symmetry of the recruited area and the asymmetry coefficients As_{DC} and As_{DR} can be influenced by a change of the ratio V_1/V_2.

4 Discussion and conclusions

In this paper we have proposed and theoretically evaluated a new method for epidural spinal cord stimulation. This method is based on a transverse tripolar electrode configuration, together with a dual-channel pulse generator. According to the simulations, stimulation with such a system is very flexible and has some important advantages over the usual monopolar or rostrocaudally arranged multipolar systems. The use of voltage sources instead of current sources has a practical advantage with respect to the field steering. For example, to achieve the asymmetry of Fig. 5, the current in generator V_2 (Fig. 3) is negative, although the voltage V_2 is zero. If V_2 is a current source, this generator has to be bidirectional to obtain the same asymmetrical field.

The primary goal of the simulations is to illustrate the changes of nerve fibre activation patterns that result from changes in stimulus parameters, such as the voltage balance of the channels, monopolar versus tripolar stimulation and simultaneous versus alternating stimulus mode; however, we do not predict exact thresholds of DC and DR fibres. We assume equal DC and DR fibre diameters and a uniform fibre diameter distribution in the entire transverse section of the dorsal columns, even though the real diameter distribution is not that simple (HOLSHEIMER *et al.*, 1991; STRUIJK *et al.*, 1992, 1993a). Therefore, the values of several evaluation parameters should not be invested with absolute significance.

The dual-channel set-up of the system provides the opportunity of two distinct operating modes: simultaneous and alternating. If the central contact of the transverse tripole is a cathode and both outer contacts are anodes, the simultaneous mode yields a recruited area that is restricted to the medial part of the dorsal columns; whereas in the alternating mode the recruited area also includes the lateral parts and, possibly, parts of the dorsolateral columns.

If the central contact is used as a cathode and the outer contacts as anodes in the simultaneous mode, the thresholds of DR fibres are relatively high compared to stimulation with a monopole and most rostrocaudal configurations. Therefore, the thresholds of radicular responses will be increased significantly, which is desirable especially when stimulating at mid-thoracic levels (BAROLAT *et al.*, 1991).

If the central contact is used as an anode and the outer contacts as cathodes, the thresholds of DR fibres will be relatively low. This may be useful if only a small part of the body, corresponding to the dorsal roots at the segmental level of the contacts, should be covered with paraesthesiae, although the probability of motor responses will also be higher.

In both operating modes it is possible to move the recruited area in a lateral direction, and consequently to affect the asymmetry coefficients for DR and DC fibres. Therefore, it is possible to change the paraesthesia coverage by changing the amplitude ratio of the two channels to allow for a difference between radiological and physiological mid-line, histological changes, lead migration etc. Lead positioning during surgery is less critical because corrections of paraesthesia coverage can be made after implantation.

The voltages required to excite DC and DR fibres are higher in transverse tripolar stimulation than in monopolar or rostrocaudal multipolar stimulation. In particular, voltages are relatively high if the thickness of the dorsal CSF layer is large. Owing to the higher voltages, the current drain is also relatively high in transverse tripolar stimulation, and therefore battery longevity will be less than in monopolar or rostrocaudal multipolar stimulation with conventional leads.

The geometry of the lead used in this study is chosen according to the criteria given in Section 2.1. This geometry can be optimised by further modelling. The optimum design of the lead may be different at various spinal levels, due to differences in spinal cord geometry and thickness of the dorsal CSF layer (HOLSHEIMER *et al.*, 1994).

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