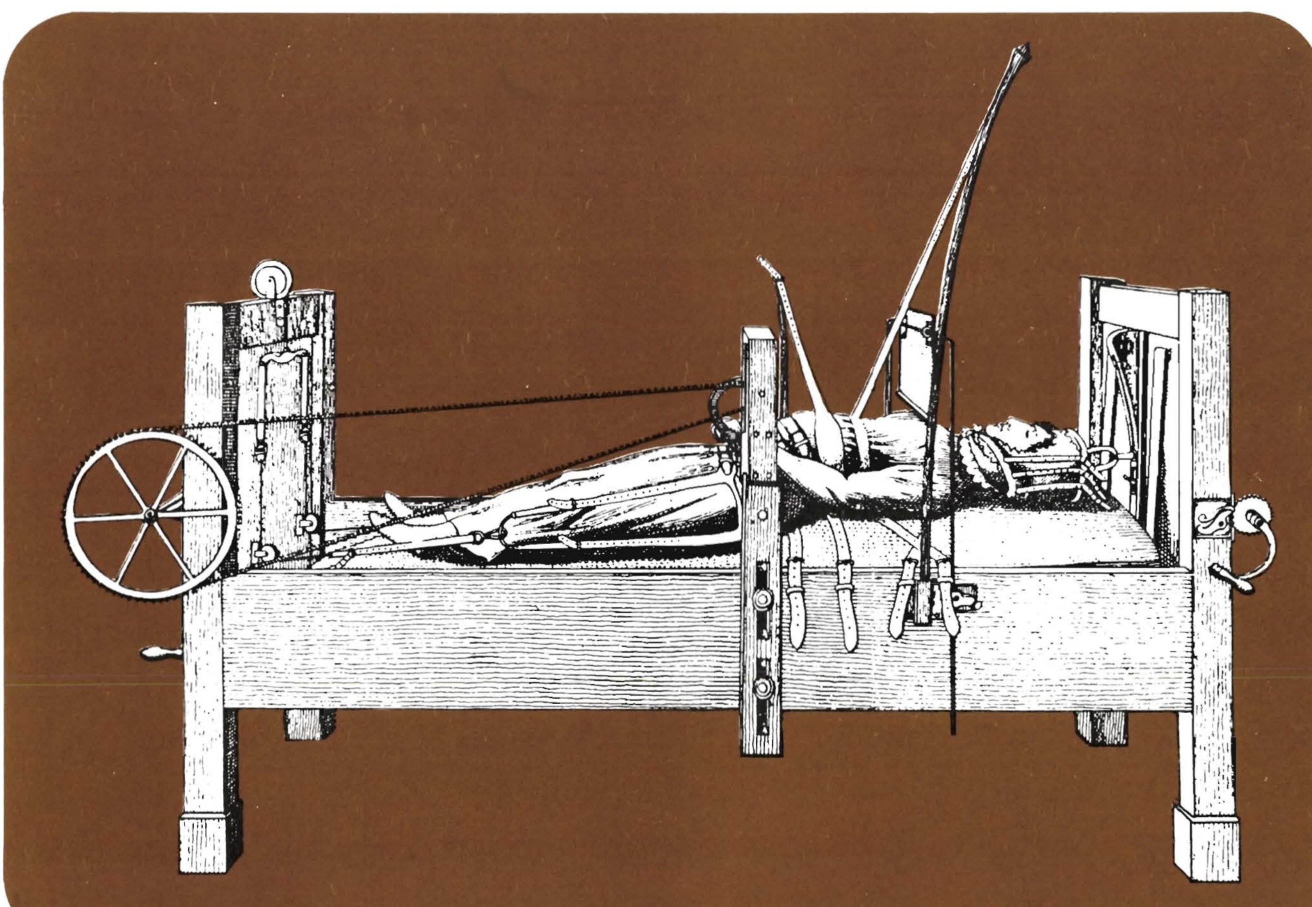
Mechanics of the spine

Biomedische Techniek



Technische Hogeschool Eindhoven / Afdeling der Werktuigbouwkunde / Vakgroep WLV

M E C H A N I C S   O F   T H E   S P I N E

Appreciation of its flexibility-rigidity, postural control and correction of the pathological spine

by

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C O N T E N T S

1. Introduction.

A mechanical construction or structure can be appreciated for its strength or for its deformation.
In the first case a comparison is made between active and admissible stresses in the material.
In the second case, which is relative to what follows, the following three aspects of the problem, which are mutually related (figure 1), are of interest: Load, Rigidity (to be derived from material properties and form, composition and areas of cross-sections), and Form (or deformation).
Given two of the three states, the third state can be determined; let us demonstrate this concept with the aid of examples associated with the spinal structure.

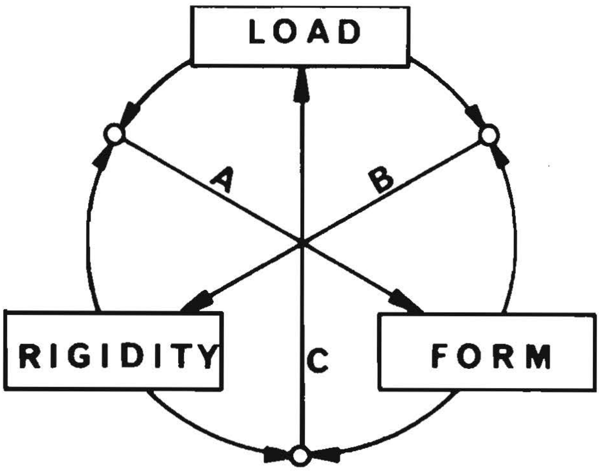


fig. 1. Scheme of the mutual relationship between form, load and rigidity. The arrows A, B, and C show the possible ways in which, by connecting two phenomena, the third one can be found.

Example A : In the case of known material properties and a known load, the deformatio that takes place can be calculated.
In a child with a flexible spine and a low bodyweight as well as in an adult with a stiffer spine and a higher bodyweight, the form of the spine will fit in as a third state. The adaptation of form could be elastic or even irreversible under certain circumstances, as in the case of scoliosis (wherein, the normal material of the vertebral bodies deforms under a load too high and asymmetrical, as shown in figure 2).

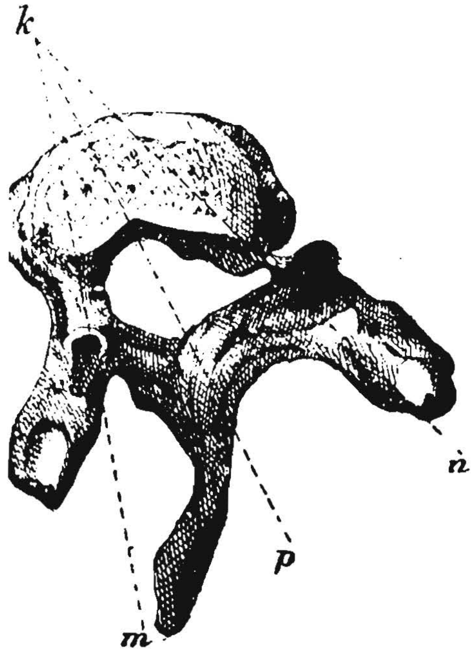


fig. 2. Deformation of a vertebra as a consequence of scoliosis

Example B : Initiating from form and load an estimate can be had of unknown material magnitudes. Thus,the load might be normal,and yet an extreme form might exist due to a degradation in material properties as in the cases of vertebrae affected by tumors or tuberculosis(fig. 3).

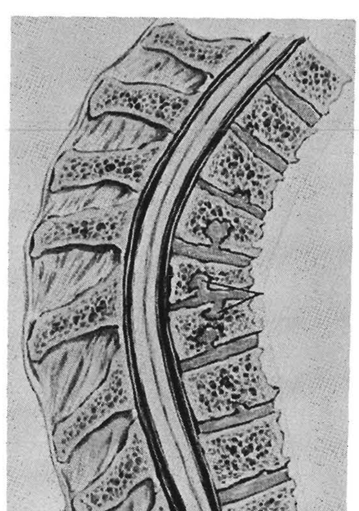


fig. 3. Deformation of a vertebra as a result of deviations of material in the case of Sheuermann's disease.

Example C : If form and material properties are known, the load can be calculated. Every deviation of form, that is not caused by deviations of material is the consequence of the load.

2. Description of form for the spine.

The form of the spine can either be determined from X-ray pictures or from the external dorsal contour. In both cases the form of the spine should be determined separately. Both methods imply specific dis- and advantages.

2.1. Form on X-ray pictures. The form of the spine is to be understood here as the smooth line passing through the geometrical centres of the vertebral bodies and of the intervertebral discs. In order to determine the geometrical centres, tangents are drawn at the ventralmost and dorsalmost limitations of the vertebral bodies; moreover, lines are drawn through the caudal and cranial limitations (fig. 4).

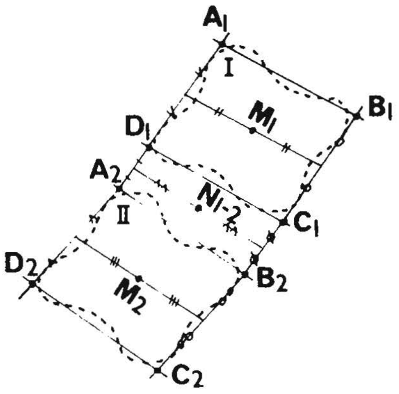


fig. 4. Determination of the geometrical centres of vertebrae and intervertebral discs.

In this way quadrangles are obtained whose geometrical centres are determined by connecting the midpoints of the opposite sides; the geometrical centres are expressed in terms of the chosen co-ordinate system. The form of the spine is obtained by connecting these geometrical centres. Figure 5 shows the form outline, in which the t-axis is oriented in the direction of gravity and the x-axis is oriented perpendicular to the t-axis in the dorsalmost point of the kyphosis, taken to be the origin (0,0); a general point C has the co-ordinates (x_c, t_c). A measuring apparatus was developed by means of which the co-ordinates can be recorded direct on a punch-tape through electronics (fig. 6).

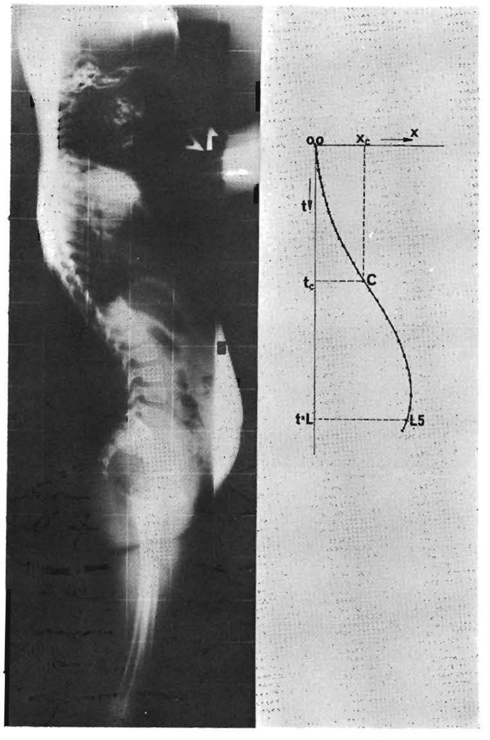


fig. 5. Description of form of a lateral X-ray picture taken in standing posture. The geometrical centres of vertebrae and intervertebral discs are marked x.
The origin of the system of axes is in the dorsalmost point of the kyphosis.

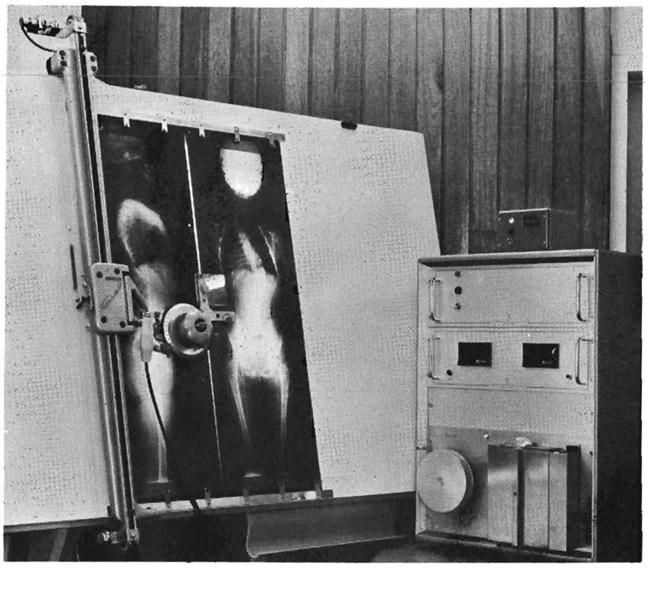


fig. 6. Apparatus developed for direct recording of geometrical data on punch tape.

The punch-tape with co-ordinates is the input-tape for a computer program for the least squares method,to fit the measured points with the curve x = F(t) of type

$$x = A.t^3 + B.t^2 + C.t + D.\sin \frac{\pi .t}{L} \qquad (1).$$

The mathematical characterisation of the form enables us to calculate the length of the curved line and the curvature at every point of the curve according to the formula:

$$K = \frac{x''}{(1 + x'^2)^{\frac{3}{2}}} \qquad (2)$$

in which $x' = 3.A.t^2 + 2.B.t + C + \frac{\pi}{L} . D.\cos \frac{\pi.t}{L} \qquad (3)$

$$x'' = 6.A.t + 2.B - \left(\frac{\pi}{L}\right)^2 . D.\sin \frac{\pi .t}{L} \qquad (4).$$

In cases of scoliosis the description of the form of the spine in Anterior-Posterior projection is also of interest. In this projection a y-axis is placed perpendicular to the X-Z plane(at the origin through the geometrical centres of the first thoracic vertebra),the formula of equation(1) is again employed by replacing x by y; a spatial system of co-ordinates is thereby obtained(see figure 7).

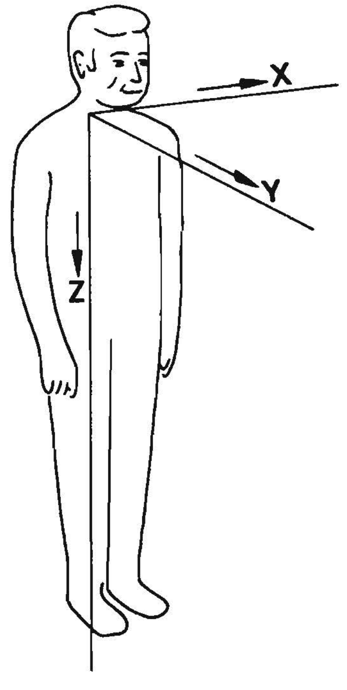


fig. 7. Spatial system of co-ordinates. The origin is situated in the geometrical centre of the first thoracic vertebra.

In figure 8 the descriptions of form in both planes of projection are shown for a scoliotic spine. By combining them mathematically a space curve of the spine is obtained; figure 9 shows its topview which is the projection on the x-y plane.

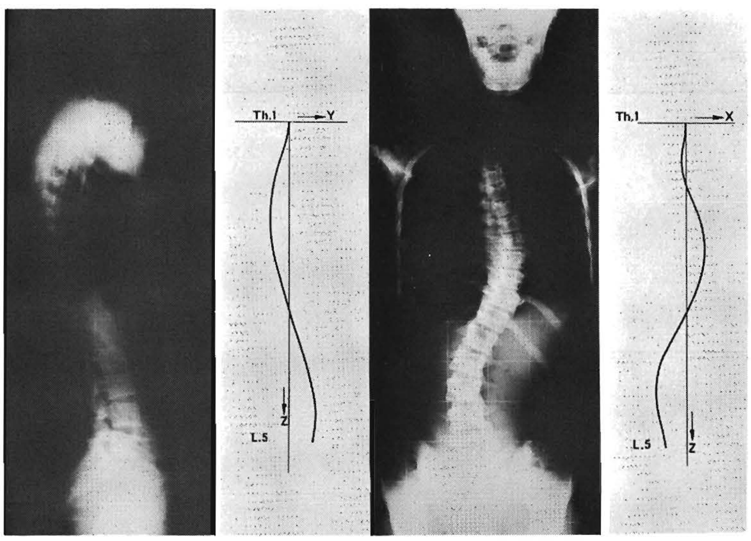


fig. 8. Descriptions of form of a lateral and an **A.P.** projection of the spine in a case of scoliosis. Prior to the description of form, a correction is made as to the proportion of the X-ray projection

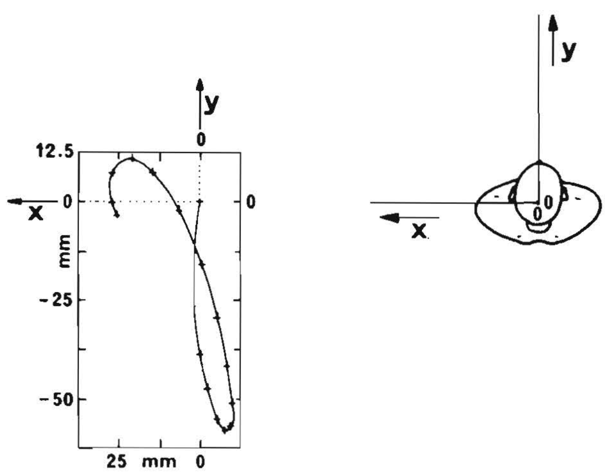


fig. 9. Top view of the spatially curved form of the spine in fig.8.

2.2. Form of the dorsal contour.

The smooth line connecting the points on the skin situated on the level of the dorsalmost points of the processus spinosi,is taken as the form of the dorsal contour of the spine.

This form cannot be recorded exactly on a lateral picture or projection because it is hidden behind i.a. the scapulae. Therefore,the back has to be approached from behind and to avoid any mechanical contact with the back,which might disturb the unconstrained posture,an optical system(reflectometer) was chosen*). By means of the apparatus shown in fig. 10,the points on the skin are recorded successively on a recording roll. By placing the subject with the heels against a wooden block,a good reference line for the posture of the spine is obtained,as the vertical through the hindmost limitation of the heels; this line represents the middle of the recording roll. In order to be certain the person stands still during the time of recording,which takes approximately 45 seconds, a light support is applied at the shoulders and at the pelvis by means of the fixator mounted in front of the person.

It was ascertained that the form and the posture of the spine can be reproduced well in case of unconstrained standing posture by ensuring reproductive positions of the feet and the hands and a similar direction of looking.

It was also observed that in normal circumstances the form of the spine remained remarkably constant during several years.

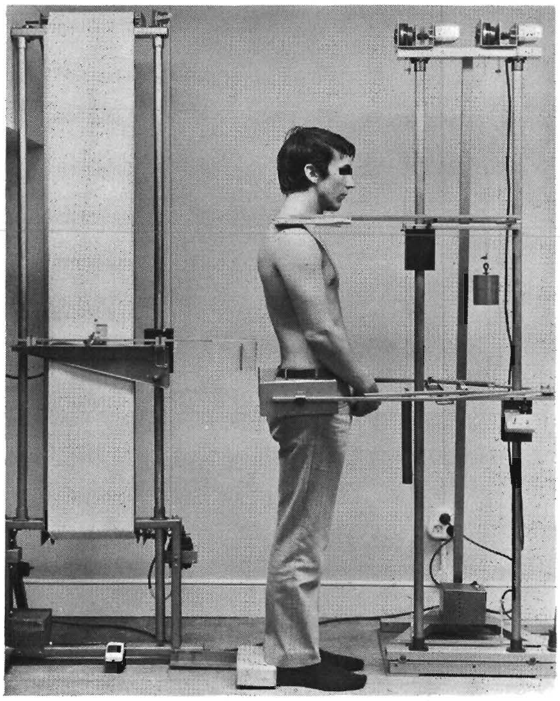


fig. 10. Recording the dorsal contour of the spine.

*) C.J.Snijders(1971): On the form of the human thoracolumbar spine and some aspects of its mechanical behaviour. Ed.Stichting V.A.M. Voorschoten,Holland.

With this method,statistical research as well as research of alterations in form in special circumstances,is very realizable.
Particularly since X-ray pictures are not needed,this method is appropriate for examining healthy people and for examination in cases in which X-ray pictures are undesirable.
In fig. 11 the spinal contours of two different subjects are shown. In both cases, the lines calculated by the computer are at most 1.7 mm. from the points measured.

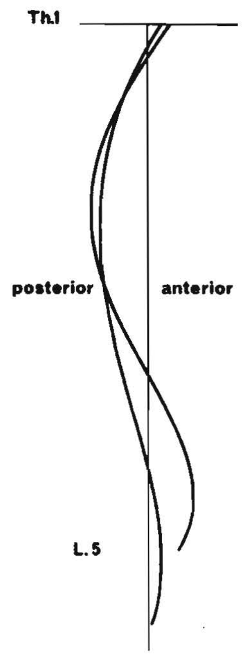


fig. 11. Description of the dorsal contour. Different forms of two persons in unconstrained standing posture. The straight line is the vertical through the hindmost limitation of the heels.

3. Rigidity,strength and material properties of the spine.

Broadly speaking the mechanical behaviour of the spine can be compared with the behaviour of an elastic rod,since its width and thickness are small in comparison with the length; its deformations due to moments of bending and of torsion are considered separately.
As the deformation to a load applied longitudinally(causing compression of the intervertebral discs) is proportionally small,this deformation is not taken into consideration.

3.1. Flexural rigidity.

The spine consists of a large number of vertebra-disc-vertebra segments(fig. 12).

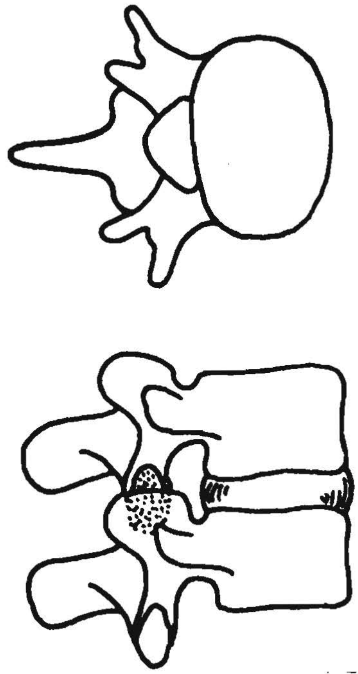


fig. 12. The spine consists of rigid vertebral elements and of slack intervertebral elements.

A model is introduced to schematize the way in which elementary force phenomena are transferred. Therein, the spine is supposed to be a continuous elastic thin-walled tube, reinforced internally by means of discs, separated from each other by a medium(with a hydrostatic behaviour and a capacity of imbibition) surrounded by a stiff layer(as shown in figure 13).
This model is able to transfer tensile forces, compression, bending and torsion.
It is however not appropriate for transfer of transverse forces, since the vertebral bodies will not remain "in line"(olisthesis).
For a satisfactory description of the transfer of great transverse forces, the help of the intervertebral joints must therefore be added.
As 17 vertebra-disc-vertebra segments exist in the thoracolumbar area, the flexural rigidity(representing the average of the rigidity of the vertebral bodies and of the intervertebral discs and the ligaments) will be expressed in units of spine length.
When characterising rigidity, we employ the following formula:

$$K - K_o = \frac{M}{E.I} \qquad (5)$$

in which: K_o = curvature in unloaded position(mm^{-1})

K = curvature after introducing the bending moment(mm^{-1})

M = bending moment(Nmm)

I = second moment of area(mm^4), derived from form and area of the cross-section at the section under consideration

and E = modulus of elasticity(Nmm^{-2}), defined by

$$\varepsilon = \frac{\sigma}{E} \qquad (6)$$

wherein ε is the strain of the material; for biological materials, E is not in proportion to σ (stress). For small changes in σ ,this relation may be considered linear.

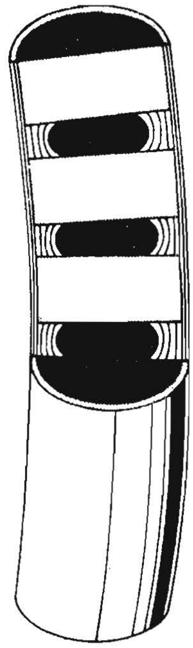


fig. 13. Model of vertebral motion segments.

When characterising the strength of the structure,the following formula is applied:

$$\sigma = \frac{M.e}{I} < \sigma_b \qquad (7)$$

in which: σ = occurring stress(Nmm^{-2})

σ_b = stress,beyond which fracture occurs(Nmm^{-2})

e = ultimate distance of material(mm)

I = second moment of area(mm^4).

The physiological requirements of the spine are

1. sufficient strength : fracture is prevented
2. sufficient rigidity : physiologically the spine must be sufficiently flexural and sufficiently elastic.

In order to gain an insight into the factors governing the construction of the human spine and appreciate spinal rigidity cum strength,the spine is compared with rods of solid bone and steel,which have the same rigidity and strength.

For a simulation of spinal rigidity,we initiate from a spine with a rigidity in ventro-flexion of E.I.= 4.10^6 Nmm^2 on the level of the third lumbar vertebra with a thickness of the vertebral body of 39 mm, measured in the sagittal plane. Further, noting that the modulus of elasticity of compact bone is $E = 183.10^2$ N/mm^2 *), it can be calculated how thick a bone rod with a round cross-section must be, to possess the same rigidity as the spine considered:

$$I_{bone\ rod} = \frac{\pi}{64} . d^4 = \frac{EI_{spine}}{E_{bone}} = \frac{4.10^6}{1,83 . 10^4} = 218\ mm^4$$

Consequently the diameter of the bone rod must be:

$$d_{bone} = 8,2\ mm.$$

A similar calculation for a steel rod with a round cross-section and $E = 2,1 . 10^5$ N/mm^2 gives a diameter $d_{steel} = 4,44$ mm(fig. 14).

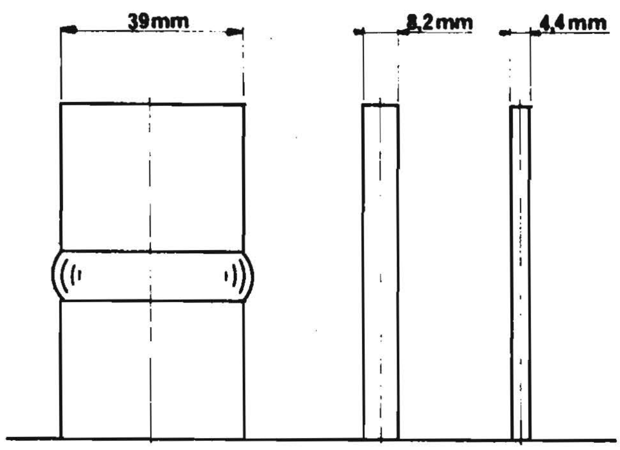


fig. 14. Comparison of spine, rod of bone and rod of steel, all having the same flexural rigidity.

For a simulation of spinal strength, it is noted that our equivalent rod of compact bone with a diameter of 8,2 mm, having a tensile strength of 140 N/mm^2 *), can only withstand a bending moment of 7578 Nmm; and for the distance of 43 mm(being the distance from centre vertebra-centre vertebra) fracture will occur for this rod of bone with a flexion of approx. $5,5^o$. However, the same length of the human spine is in a position to transfer a bending moment of at least twice as large(about 16000 Nmm) for which bending moment a bone rod would need a diameter of at least 10,3 mm.

... of biological materials. Williams and Wilkins

However, the disadvantage of this bone rod is the rigidity which is at least 2,5 times as large as the rigidity of the human spine.
At the same time, in order to allow double bending, the bone rod would have to be twice as thin (3,25 mm).

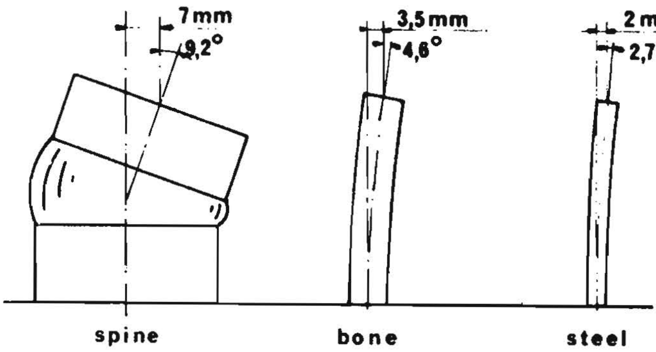


fig. 15. The rods of bone and steel with the same flexural rigidity as the spine, are, compared with the spine, limited in their ventro-flexural movement by their limitations of breaking strength (or limitations of elasticity).

Further, considering that spongy bone has a tensile strength that is less than a tenth of compact bone, it can be posited that a bone rod cannot give the combined requirements of rigidity and bending strength (sufficient bending strength also guarantees sufficient compression strength).

3.2. Material properties.

In order to relate the load to the form, it is necessary to know the material properties.
As the flexibility of the spine is important to us, in this chapter, we will deal with the modulus of elasticity and the second moment of area; the breaking strength, creep in structures and the elasticity of the parts of the spine, like bone, ligaments, disc-tissue etc., are omitted.
We are taken into consideration the elasticity of the spine as a unit. The following way of calculation by which the elasticity of spines can be calculated roughly, cannot be exact; individual structural differences will exist. We are only interested in indications in certain directions, evaluating matters concerning laws in order to gain an insight into representing the elastic behaviour of the combination: stiff vertebral body - flexible disc.

In order to have quantitative information about the modulii of elasticity and shear of the spine, bending and torsion-experiments have been made with autopsy specimens. The measuring instruments shown in fig.16 has been used for the said experiments.

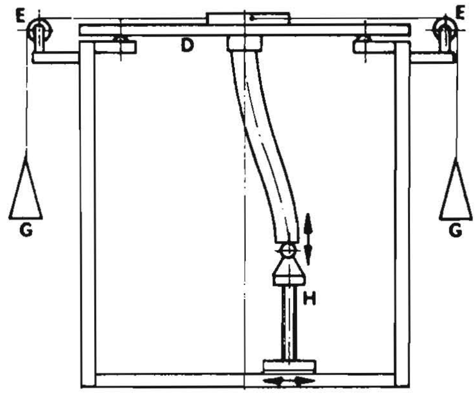


fig. 16. Schematic representation of a spine-loading instrument.

In this instrument the topmost point of fixation of the spine can translate freely, when the spine is loaded in torsion. The clamp (D), between which the topmost vertebra is fixed, is connected to a large perspex disc that has been placed on roll-bearings.

In order to load the spine with a torsional moment, equal weights are placed on the pans G. In order to load the spine with a bending moment, the spine is clasped at H and the glass disc is taken away; one of the wires is now directly attached to the topmost spine-clamp, and by placing weights on the corresponding pan G, a bending moment is imposed that is graphically represented in fig. 17.
In order to determine the bending rigidity of the spine, nails are driven into the vertebral bodies (as shown in fig. 18) to serve as reference lines. From the mutual angular rotation (Ψ) of these lines (at various loads), the rigidity of the spine can be obtained.

$$(EI)_d = M_b/(\Psi/d) \qquad (8)$$

Since it is the flexible disc (rather than the stiff vertebral body) that facilitates rotation, the stiffness $(EI)_d$ assessed with respect to the disc thickness d, is more amenable to measurement than $(EI)_l$; the latter can however be obtained from $(EI)_d$ with the help of the following relation

$$(EI)_l = (EI)_d\ (l/d) \qquad (9).$$

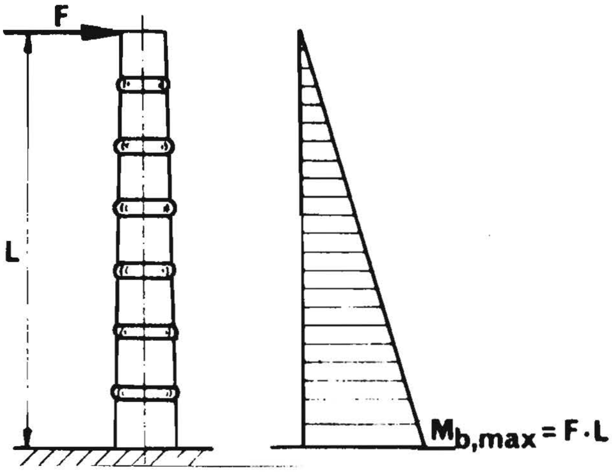


fig. 17. Bending moment during the experiments of loading.

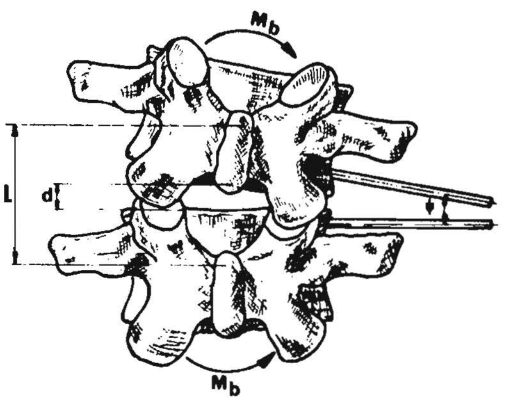


fig. 18. Relation between applied bending moment and mutual angular rotation.

We see that $(EI)_d$ can be calculated from the experiments,but we should like to know the modulus(E). As a first approach we assume that the second moment of area of the cross section of the elliptical disc is proportionately represented as

$$I \propto a^3 b \qquad (10)$$

in which the dimension 'a' is the diameter of the vertebral body concerned,lying in the plane in which the spine is bent,and the dimension 'b' is the diameter perpendicular to this plane. When $(EI)_d$ is divided by $(a^3 b)$, a material constant E' is obtained,which is proportional to E,the modulus of elasticity of the disc and ligaments. In fig. 19 the spine is visible in loaded and unloaded positions,if bent forward and bent laterally.

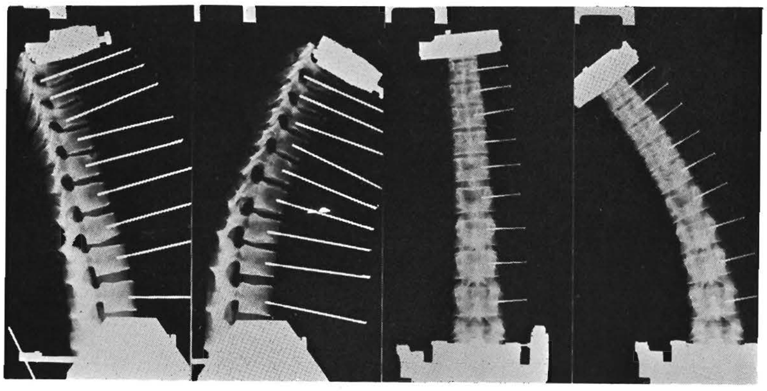


fig. 19. X-ray pictures of the load-experiments before and after load.
Left: bent-forward position.
Right:bent-laterally position.
(female,aged 18).

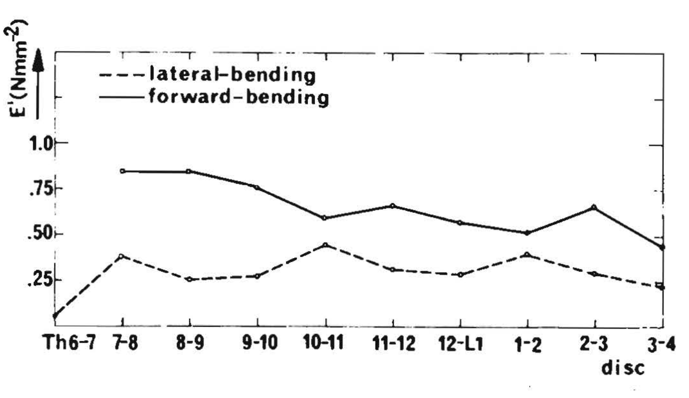


fig.20. E'-values for the spine of fig.21. Approximately linear for the zero load upto the maximum .(female,aged 18).

In fig. 20 the calculated E''s(along the spine) have been drawn graphically(for both forward and lateral bending experiments).
The stiffness $(EI)_l$ can be obtained from the E' as follows:

$$(EI)_l = E'(a^3b)\left(\frac{1}{d}\right) \qquad (11)$$

wherein

E' = constant of elasticity from fig. 20

a = vertebral diameter in plane of bending

b = vertebral diameter perpendicular to plane of bending

l = distance intervertebral disc-intervertebral disc.

d = height of disc.

Thus, by employing the E' of figure 20 and by determining the dimensions(a,b,l & d) from X-ray pictures(after applying appropriate correction factors for X-ray projection distortion),we can obtain the in vivo value of the rigidity of the spine.

The E' given in fig. 20 is measured from a not pathological spine with the age of 18. Thus when we use formula 11,to get an impression of the rigidity of a particular patient with the help of his X-rays,we get a non-pathological and "not degenerated" rigidity. When the patient has specific pathological changes on his spine or when the patient has an age at which the discs had degenerated other values then the E''s of fig. 20 must be used.

In order to determine the torsional rigidity of the spine,we apply the same method as when calculating $(EI)_l$ and employ the formula

$$(GI_p) = G' \frac{a^3 b^3}{a^2 + b^2} \frac{1}{d} \qquad (12)$$

wherein G = modulus of shear(Nmm^2)

I_p= second polar moment of area(mm^4).

The experimental setup for the torsion experiment with clasped spine is shown in fig. 21.

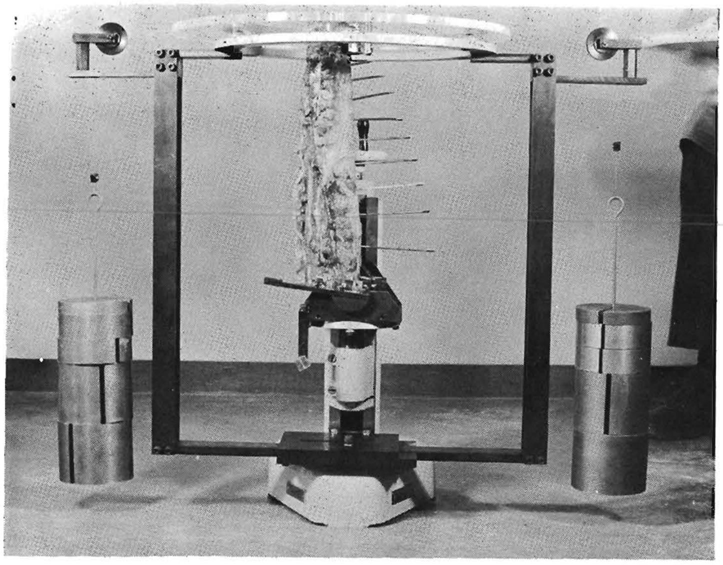


fig. 21. The torsion test.

The orientations of the vertebral bodies relative to one another are designated by nails driven into them. By taking photographs from above,the angular rotations of these nails at the various loads can be measured(fig. 22). The values of G' calculated with the help of these photographs have been drawn in fig. 23.

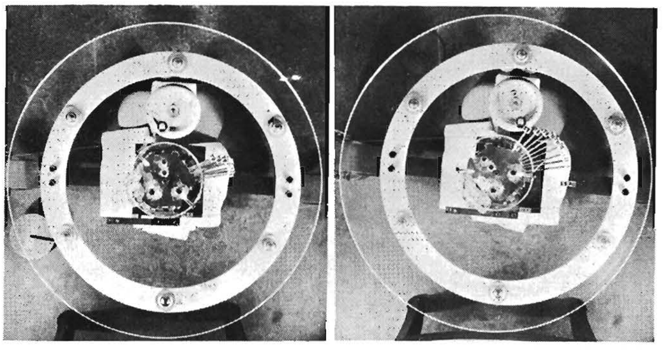


fig. 22. Photographs, taken from above, of the spine in an unloaded, resp. loaded position.

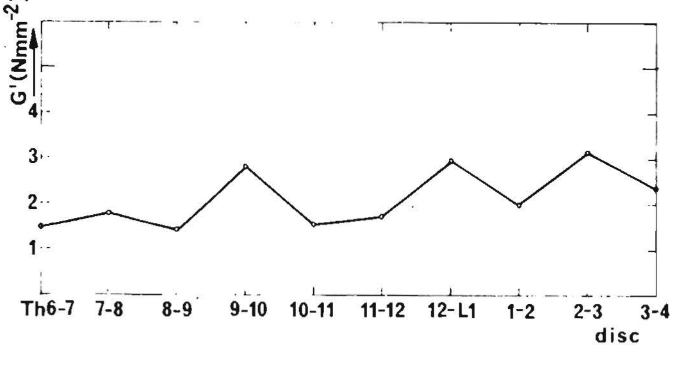


fig. 23. G'-values of the spine of fig. 21.

From the aforesaid approach, posited in the formulae (11) and (12), we see that a spine is more rigid for (i) a larger vertebral diameter and (ii) for a greater value of(l/d), the quotient of distance between centres of successive vertebrae and height of disc, for not-pathological cases.

4. Determination of the form or the load-bearing capacity of the spine.

In order to determine the form from the load and material properties, we employ

$$K = \frac{M}{EI} \qquad (13)$$

in which

K = the curvature of the material in a certain plane

M = the bending moment exerted in that plane

E.I = the flexural rigidity of the cross section considered.

In employing the above formula, it is implied that the influence of the transverse forces on the deformation is negligible. Alternately, it is more pertinent to measure the form parameters and, with the help of known material properties, calculate the mechanical load of the spine responsible for producing the in vivo form. For this purpose, we employ the above equation to determine the bending moment

$$M = K.E.I \qquad (14)$$

wherein K equals curvature of the deformed spine if it is straight in the unloaded state and equals change in curvature if the spine is curved even in the unloaded state, as in the case of scoliosis.

We, of course, need to know the mathematical description of form of the spine (fig.24).

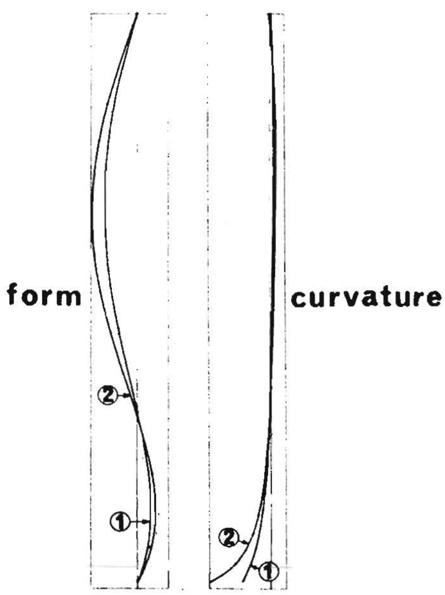


fig. 24. From the description of form, the curvature at any level can be calculated immediately.

The determination of the form function x = x(t) and the calculation of the curvature K therefrom, has been dealt with in section 2. From equations (14) & (12), the bending moments are obtainable (in vivo) from the following expression

$$M(t) = \frac{x''(t)}{(1 + x'(t)^2)^{\frac{3}{2}}} \cdot E'.a^3.b.\frac{1}{d} \qquad (15).$$

5. Scale factors or non-dimensional parameters for comparisons of spines or skeletons as mechanical structures.

We have said earlier that a mechanical structure or a mechanical construction can be

appreciated for its load carrying capacity or for its form or deformed shape; of course each property is related to the other by means of the rigidity of the structure. In order to compare the effectiveness or efficiency of skeletal structures, we need to obtain non-dimensional parameters that incorporate both the load-bearing and form characterising properties of the structure.
Alternately we can compare either the strengths or stresses of structures in terms of geometrical dimensions. Thus the stresses due to self-weights of two structures of the same density are proportional to their characteristic lengths(L),since the weight is proportional to the cube of the characteristic length and the cross-sectional area is proportional to the square of the characteristic length.

Examining grown-up animals of the same form,we see that with the smaller ones the skeleton constitutes a smaller percentage of the weight.
The explanation is as follows: the muscular forces are in proportion with L^2 and the forces due to self-weight are proportionate to L^3. This means that the forces on the skeleton(due to self-weight) increase more than proportionally to the height or length at a faster rate than the muscle forces.
As a consequence(in case the material properties remain the same),the loaded cross-sections must increase more than proportionally;in other words, a bigger animal will have a thicker skeleton than a smaller animal would have,if it were to have the same length as the bigger one. Figure 25 illustrates this point graphically.

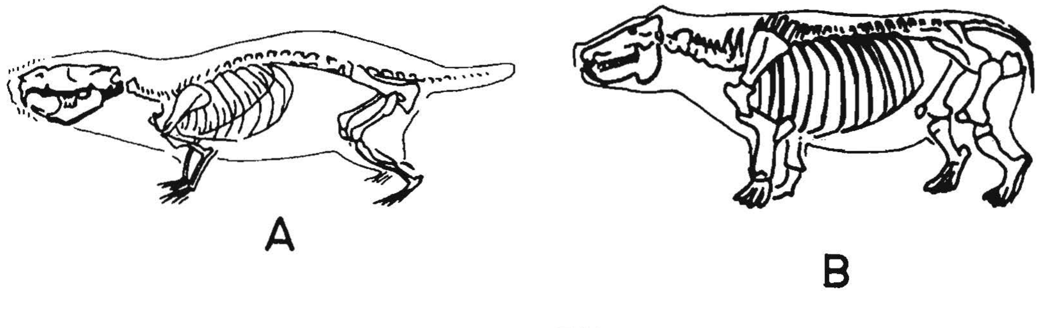


fig. 25. Skeletons of (A) a lemming and (B) a hippopotamus,both reduced to a same bodylength,to show the much greater robustness of the skeleton of the hippopotamus. Hence the troubles of bigger animals due to their own weight,are ever-increasing(Hesse-Doflein,1935).

Let us now study & determine the governing criterion for skeletal growth of man.
One has to keep in mind that the construction material of the child is different from that of the adult;the child's skeleton for the greater part consists of cartilage whereas the adult's skeleton consists of hard bone.

Let us test a hypothesis that the process of growth takes place in such a way that the stresses in the bearing elements remain constant for increasing weight. In order to test this hypothesis we take a bone cylinder of dimensions h and d(fig.26),loaded by a bending stress as a consequence of the uppertorso weight W.

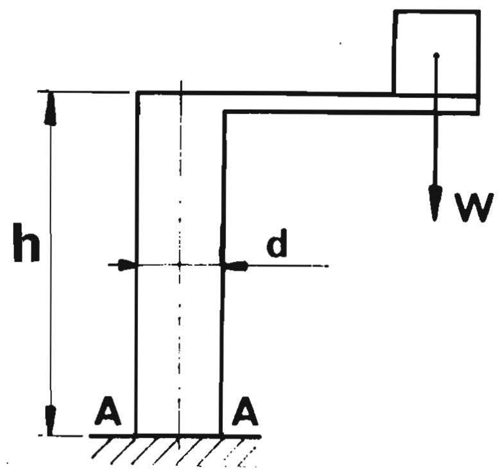


fig. 26. **A part of the skeleton is loaded with the weight of the upper part.**

The bending moment M in cross section A - A equals the weight W multiplied by the lever arm. This arm is proportionate to the width dimension d $\rightarrow$ $M \propto W.d.$

The weight W of the upper part of the skeleton equals the volume of this part multiplied by the specific mass, ρ . The volume is proportionate to $d^2.h.$

Thus, $W \propto \rho.d^2.h.$

$$M = \rho.d^2.h.d = \rho.d^3.h.$$

The bending stress in cross section A - A is

$$\sigma = \frac{M}{Z}$$

wherein Z is the moment of resistance of the cross section and is proportionate to d

Thus, $\sigma = \frac{\rho.d^3.h}{d^3} = \rho.h$ (16)

The bending stress σ consequently depends on the height h in the direction of gravity and does not depend on the diameter d.

Assuming that the stresses in the bearing elements during growth remain constant, this implies inevitably that the height of length h is not allowed to increase(provided of course that the material density remains constant),whereas width and thick-

ness may increase.
However, the height of man does increase, which inevitably means that the stresses also increase, thereby violating our hypothesis of constant stress.

Another criterion that can be tested is that of the strains remaining the same.

If the strain(ϵ) has to remain constant, the quantity σ/E also has to remain constant, for $\epsilon = \sigma/E$.
However, since we have seen that stress(σ) increases, E must also increase.
This implies that the bone must grow more rigid and stronger.
The admissible stress and the modulus of elasticity of the bone, taken as a composite of hydroxyapatite and collagen, are given by

$$\sigma_c = \sigma_H \cdot V_H + \sigma_M \cdot (1 - V_H)$$
$$E_c = E_H \cdot V_H + E_M \cdot (1 - V_H) \qquad (17)$$

in which σ_c, E_c = admissible stress, modulus of elasticity of the composite

σ_H, E_H = admissible stress, modulus of elasticity of the hydroxyapatite

σ_M, E_M = admissible stress, modulus of elasticity of the matrix collagen

V_H = degree of filling, this is the proportion of the area of the hydroxyapatite to the entire area of the composite in a cross section perpendicular to the axis of the bone.

The total strain,

$$\epsilon_c = \frac{\sigma_H \cdot V_H + \sigma_M \cdot (1 - V_H)}{E_H \cdot V_H + E_M \cdot (1 - V_H)} = \frac{\sigma_c}{E_c} \qquad (18)$$

thus depending on the degree of filling, will have to adjust itself in such a way (in order to maintain constancy) that the right combination between stress and E is found. In case such a mechanism is effective in the human body, there must exist an element in the bone, recording the strain and emitting a signal to an element that increases the degree of filling; as a result, the strain returns to its original value. It is plausible that this process occurs because piezo-electric phenomena in the bone have been detected by several researchers, emitting an electric signal when being loaded(or strained). Thus the hypothesis that growth occurs according to consistent strains is probable.
Let us finally test the hypothesis that in the case of similar loads the bone-elements bend into identical forms. Consequently in fig. 27

$$\phi = \frac{M.h}{E.I}$$

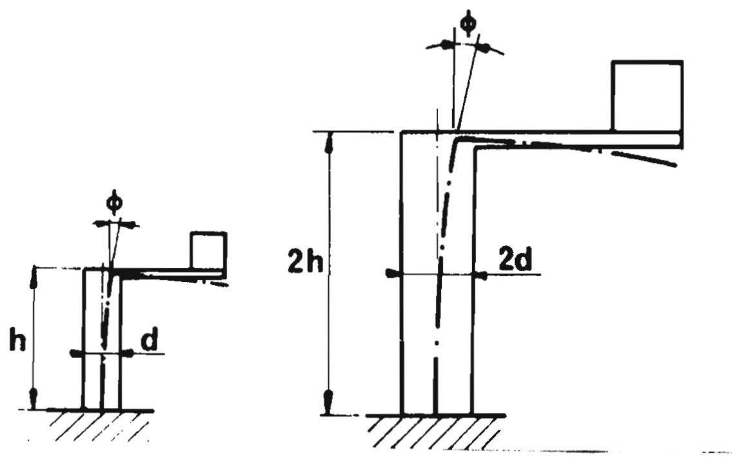


fig. 27. Deformed spine with the uppertorso weight acting at height h.

According to (16) M was $\propto \rho . d^3 . h$)
$E \propto \sigma \propto \rho . h$)
$I \propto d^4$)

$$\longrightarrow \quad \phi = \frac{\rho . d^3 . h . h}{\rho . h . d^4} = \frac{h}{d} \qquad (19)$$

Furthermore the angular rotations as to big and small constructions are the same. Thus,

$$\phi = \text{constant} \longrightarrow$$

$$\frac{h}{d} = \text{constant} \longrightarrow h \propto d \qquad (20)$$

This means that the growth in thickness of the parts of the skeleton increases proportionately to the growth in length.
In this context the volume of the skeleton is

$$V \propto d^2 . h \propto h^3$$

and consequently it is proportionate to h^3

The weight of the skeleton $G \propto \rho . h^3$
We had seen that during growth the bone structurally changes under the influence of the degree of filling. For this reason we may not suppose ρ being constant. Adapting the weight of the child mathematically to its length in the expression

$$G \propto h^n$$

and, for simplicity's sake, wanting to express n in real numbers, we find that n is quite near 4. Written as a formula we get

$$\text{for boys}: \quad G = 56{,}0 \cdot h^4 + 102{,}0 \text{ Newton}$$
$$\text{for girls}: \quad G = 59{,}8 \cdot h^4 + 93{,}1 \text{ Newton} \qquad (h \text{ in meters}) \quad (21)$$

Figure 28 shows drawn lines, representing adaptations according to formula (21); the + 's and o's are the measured values for children between 2 and 17 years of age. From this adaptation it appears that taking into consideration a reasonable approximation: $G \propto h^4$.

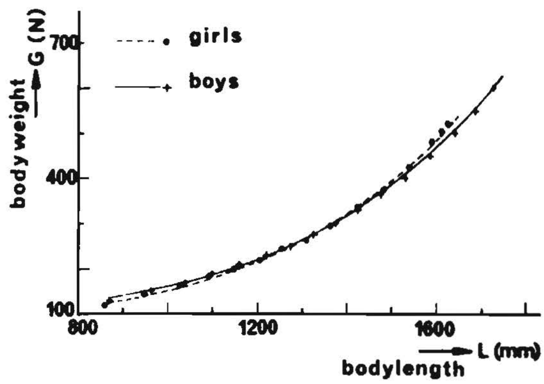


fig. 28. The weights of children between 2 and 17 years of age connected with these children's lengths.
It appears that the drawn lines according to formula (21) are adapting well.

Consequently the specific mass of the skeleton must be proportional to h.

Thus, <u>for adults the bone must not only be stronger and more rigid, but, moreover, it must also have a larger specific mass than in case of children.</u>

In order to verify $G \propto h^4$ in a different way, applying (20), this formula can also be written:

$$G \propto d^4.$$

To this effect we measured at the same level the vertebral diameters with the help of X-ray pictures of a number of persons. After this we drew these measurements in connection with the weight of the persons, and as a matter of fact a good linear relation appears to exist (fig. 29).

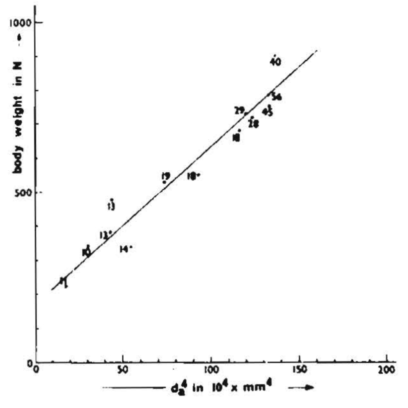


fig. 29. The weights of a number of persons drawn in connection with the fourth power of the corresponding vertebral diameters,

The curvature of the parts of the skeleton, K, is proportionate to $\frac{\Phi}{L}$,

and because Φ is constant, is:

$$K \propto \frac{1}{L}.$$

Consequently the spine of the child(small L) will be curved more than the spine of the adult(fig. 30).

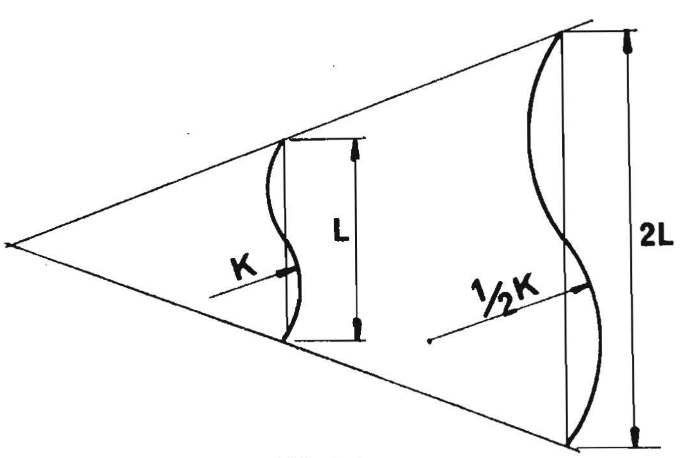


fig. 30. Regular increase as to form results in less curvature.

6. Standing and sitting as a mechanical problem.

6.1 Load on the spine.

The load affecting the spine can be divided into three kinds:

a. the load as a consequence of one's own weight

b. " " " " " " the affecting muscles and ligaments

c. the external load such as the extra-weight when lifting, the accelerating forces, etc.

The percentage weights of the various parts of the skeleton *) are schematically shown in fig. 31.

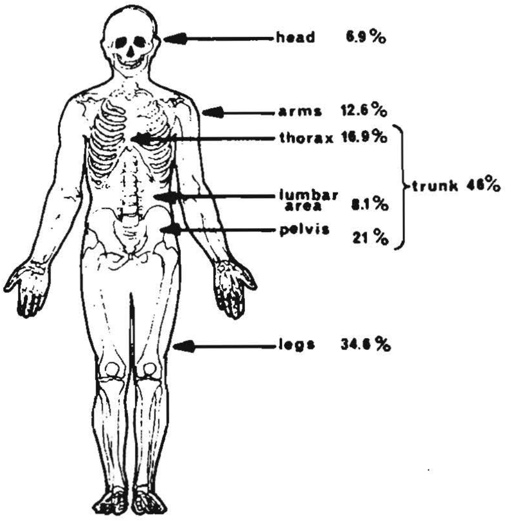


fig. 31. The division into percentages of the parts of the body.

The load as a consequence of the affecting muscles is unknown; a good coupling between the measured electro-myogram and the occurring muscular force has not proved to be possible thus far to yield this information. On the other hand we can often readily calculate the magnitude of the unknown muscular force with the help of a simple vectordiagram, provided the direction of this force is known(fig. 32).
When dealing with the influence of the load on the posture, we omit the dynamical forces and only consider the posture in unconstrained standing-position. In this unconstrained standing posture only a few groups of muscles are active and regulate the equilibrium with minimal exertion around the joints, such as ankle, knee and hip-joint.

*) Williams, M., Lissner, H.R., (1962): Biomechanics of human motion. W.B.Saunders Co.

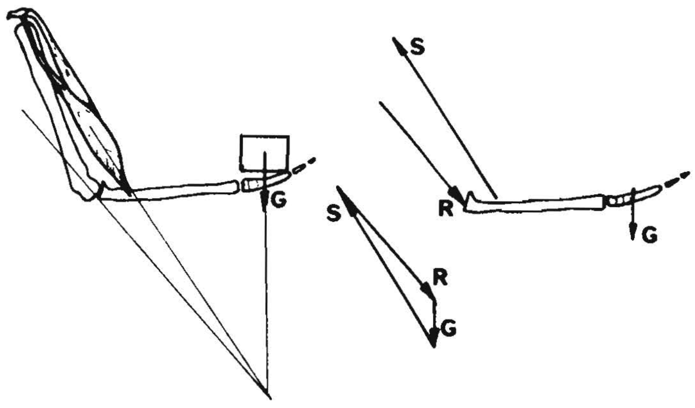


fig. 32. From a simple vectordiagram the unknown muscular force can be calculated,provided the direction of it is known.
(For simplicity's sake the weight of the arm is omitted here with respect to the weight on the hand).

The collective term for these muscles which regulate standing is Postural muscles, as opposed to the Phasic musculature which causes the movements and reflex-movements. The phasic musculature has another function and consequently a different behaviour and structure than the postural muscles. In order to gain an insight into the load on the spine and the forces acting on it due to the postural muscles,we give the following example illustrated by figure 33. Therein A is a point at which the bending moment in the spine is zero. F is the weight of the part that lies above A. Point B is the transition from the lumbar vertebrae to the sacrum and thus is the clasp in the pelvis. The load q represents the equally divided weight of the part of the spine between A and B. At point B the clasp-reactions have been drawn. The reaction force R provides vertical equilibrium; the clasping moment M can be calculated from the equilibrium of moments,and is caused at B by muscles;the most important muscle causing the moment at B is the Musculus Psoas(fig. 34).

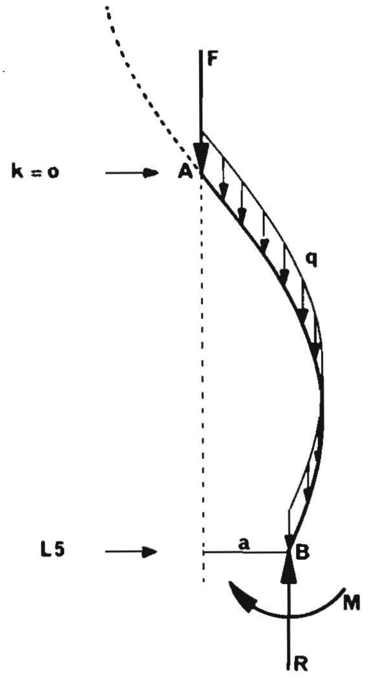


fig. 33. Forces and moments affecting a part of the spine

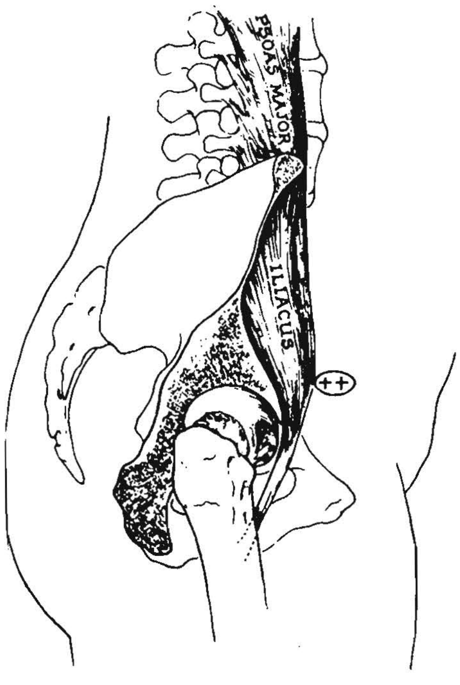


fig. 34. In erect posture the Musculus Psoas is activated positively(Basmajian,1967).

Basmajian *) monitored its activity during standing. Whereas with the quadrupeds its function is phasic,with the erect-going man it is postural and procures the moment of equilibrium around the pelvis. As a result,it is also the cause of lumbar lordosis; it can immediately be seen that when activating this M. Psoas more, the lumbar lordosis increases.

Consequently,with people having an increased lordosis the cause of it mostly must

*) Basmajian, J.V., (1967): Muscles alive; Their function revealed by electromyography. Williams and Wilkins Co. Baltimore.

be sought in connection with the existence of a shortened or slightly hypertonic Musculus Psoas.

6.2 Mechanics of standing.

In order to systematically build up a mechanical model of the skeleton,with its influences of forces,we start from a model which only contains the principal joints. These are: 1. the ankle-joint; 2. the hip-joint, and 3. the lumbar spine which is a complex of joints.

The displacements of the lumbar spine will be characterized by its curvature,whereas the displacements at the ankle & knee joints will be characterized by rotations. The curvature of the lumbar spine is the consequence of the rotations in every lumbar joint(fig. 35). The knee-joint is considered to be "locked" in the frontal plane.

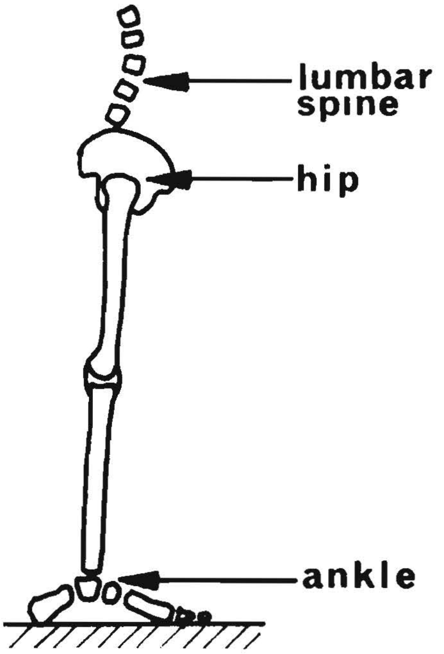


fig. 35. The principal joints relating to standing.

The method,to be applied,aims at gaining an insight into the posture as a whole and into the relations contributing to the equilibrium around voluntary joints.

Our analysis starts at the ankle-joint. When analysing equilibrium around this joint,we cut the ankle-joint imaginarily and build a free-body diagram of the foot(fig. 36).

The ankle-joint can be considered as a hinge that is able to take up forces in a horizontal and a vertical direction only and no moments.

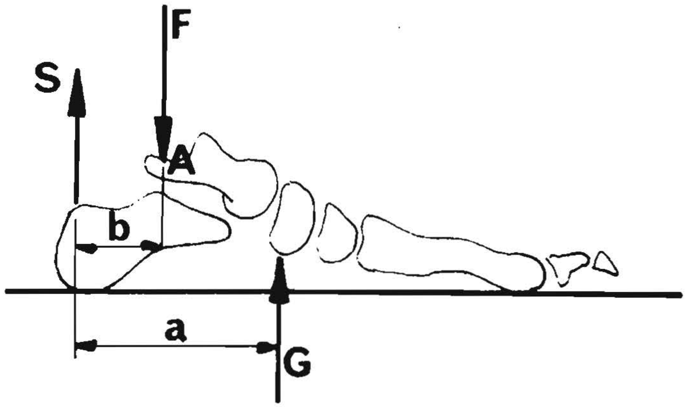


fig. 36. The equilibrium around the ankle-joint.
For simplicity's sake S is supposed to be parallel with G.

The position and the magnitude of the ground reaction G(= ½ the weight) can be determined by a **stabilograph *)**.

Considering the equilibrium of moments around **A**, we find that

$$b.S=(a-b).G \qquad (22)$$

from which the force S in the Triceps Surae group is given by

$$S = \frac{a-b}{b} . G \qquad (23)$$

For an arbitrarely choosen example with: G = 380 N., a = 115 mm., b = 60 mm., we obtain S = 347 N. Thus, when standing symmetrically, the Triceps Surae will have to be tightened to 347 N. for each leg, in order to guarantee the equilibrium around the ankle-joint. The force of reaction in the ankle-joint is obtained from vertical equilibrium considerations as :

$$F= S + G = 347 + 380 = 727 \text{ N.} \qquad (24)$$

In this example, the load on one ankle-joint has, when standing in unconstrained posture, the same order of magnitude as the body-weight. When inclining forward (in which case the magnitude of 'a' increases, this load becomes still bigger.

As to the equilibrium around the head of the hip, we consider the complete leg(which is cut loose from the pelvis imaginarily), put down and analyse the forces acting on it(figure 37).

***) C.J.Snijders, M.Verduin (1973): Stabilograph, an accurate instrument for sciences interested in postural equilibrium. Agressologie 1973, 14, C.**

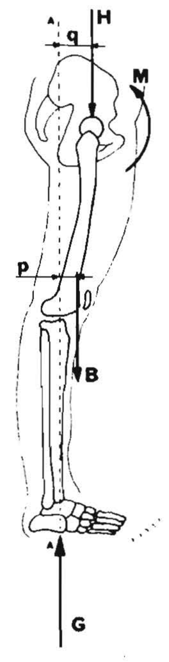


fig. 37. Forces and moments on the imaginative cut loose leg.

Figure 37 can be compared with an X-ray picture of the leg in combination with a stabilogram. Therein, G is half the body-weight, B is the weight of one leg acting through its mass centre, H is the reaction force on the head of the femur, M is the moment of reaction around the head of the femur, and the distances p and q are the distances between the vertical AA through the mass centre of the body and the lines of action of the forces B and H.

From anatomical considerations, we have

$$B = 0{,}17 \cdot G$$

The equilibrium of forces in fig. 37 has the consequence

$$H = G - 0{,}17 \cdot G = 0{,}83 \cdot G \qquad (25)$$

The equilibrium of moments around the head of the femur has the consequence

$$M = q.G - (q - p).B \qquad (26)$$

Combining equations (25) & (26), we obtain

$$M = q.H + \frac{0{,}17}{0{,}83}.p.H$$

This can be written:

$$M = e.H \qquad (27)$$

wherein $e = q + \frac{0{,}17}{0{,}83} \cdot p$ (28).

As the reaction force on the acetabulum(not to mention the direction) are the same as these on the head of the femur, we can draw the equilibrium consideration of the pelvis readily in figure 38, whence we obtain M = e.H.

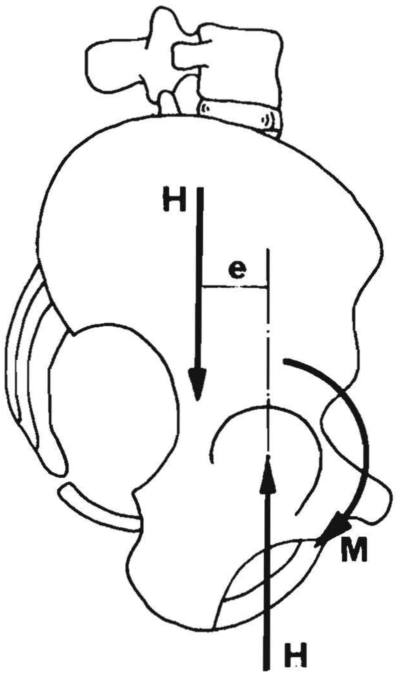


fig. 38. Forces and equilibrium of forces and of moments of the pelvis.

Another consequence is that,should the reaction force H lie behind the acetabulum, the moment around this acetabulum is clockwise. The muscles contributing to the moment around the acetabulum are as follows:

A.those causing in fig. 38 a clockwise moment: M. rectus femoris, M. psoas major, M iliacus, M pectineus, M. gracilis, M. adductor longus.

B.those causing in fig. 38 an anticlockwise moment: M. glutaeus maximus, M. adductor magnus, M. adductor brevis, MM.ischiocrural.

Both these types of muscle groups lie between pelvis and femur and take care of the posture of the pelvis with respect to the femur. When standing unconstrained a certain adjustment of these muscle groups around the hip-joint occurs. Collectively these muscles produce a moment M given by

$$M = e.H.$$

When inclining backward e increases. To provide equilibrium, M should increase proportionally; to this effect the muscle group A should be activated to a larger extent,so that the increased associated muscular strength F(having its line of activities at p,see figure 39) procures an increased moment.

When standing unconstrainedly,the distance e (figure 38) is manipulated so that M lies in an appropriate range of control.

Imagine that,in an unconstrained standing posture, H runs exactly through the hip-joint,then the following happens. Then the value of e = 0 and M = 0,too.

A slight deviation from the equilibrium results in a small value for e, say a positive value.

When e reaches a certain threshold value,the muscles around the hip-joint react & provide an opposite moment - M threshold. After having attained this threshold value,

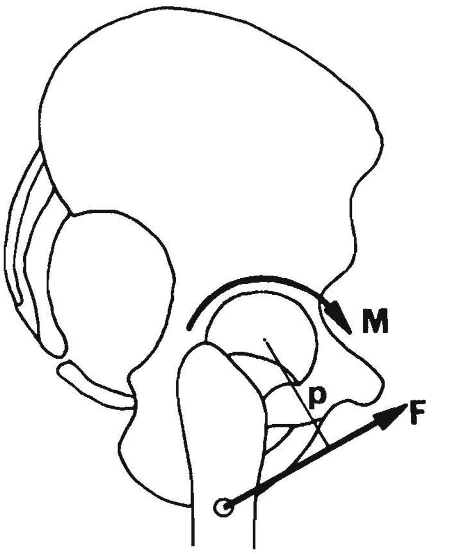


fig.39. Equilibrium of moments around the head of the femur. The diagram is incomplete with respect to the equilibrium of forces.

the muscular group F pulls back again on the head of the femur; e now becomes negative as M becomes anticlockwise. At this point the muscular groups acting to the left are tightened to again decrease 'e'.
We thus see that postural control is in this case effected with H shifting and making e oscillate from positive to negative value. In conjunction with this oscillation, an oscillatory rotation in the hip joint occurs which can result in an early wear of the hip-joint (e.g. cox-arthrosis).
The condition for this early wear consequently is: average e = 0, or according to equation (28) for two legs:

$$q + \frac{0{,}35}{0{,}65} \cdot p = 0 \qquad (29)$$

The postural control around the hip-joint entails continuous switching-over from a clockwise moment to an anticlockwise moment and in turn switching-over from muscular group a to muscular group b, which contributes to fatigue.
For this reason, when standing unconstrainedly, the postural muscles will rather be used than the phasic ones, since the postural muscles do not get tired as quickly and also the irritability is less; moreover, these muscles are stronger and have a better supply of blood. This means that group a, of which the M.Psoas and M.Rectus femoris are explicit postural, is continuously active and that, according to fig. 38, H lies constantly behind the head of the femur.

For a good posture, e is continuously positive(positive to the left) and the adjustment then takes place between $e - \Delta e$ and $e + \Delta e$. The characteristics of the postural muscles are that a small change in activation can effect a big change in force immediately.
When e increases this can immediately be compensated by a small change in the activation <u>without effecting a rotation in the hip-joint.</u>
The muscles constantly yield a tensile force around the hip-joint and arrange e in an equilibrium. This tensile force is the consequence of an activation of these muscles.

When standing unconstrainedly in a certain posture, the length of the muscle is almost constant, the degree of activation will be optimum for that specific muscle, so that it will not get tired soon. Moreover, it can adjust well around this optimum activation, the corresponding optimum tensile force is called $F_{opt.}$. The moment around the hip-joint will therefore be(fig. 40):

$$M = p \cdot F_{opt.}$$

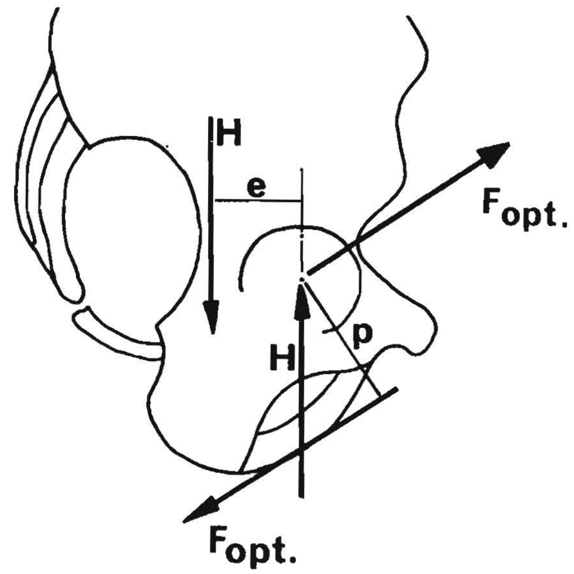


fig. 40. Equilibrium of moments around the acetabulum.

In fact the above formula must read:

$$M = P_1 \cdot F_{1opt} + P_2 \cdot F_{2opt} + P_3 \cdot F_{3opt} + \ \ldots\ldots$$

in which F_{1opt}, F_{2opt} ....etc. are the various contributing optimum muscular forces. All these forces can be combined to a resulting F_{opt} that runs at a distance p from the head of the femur.
In fig. 40, the equilibrium of the pelvis has been drawn; we can observe, therefrom,

$$e \cdot H = p \cdot F_{opt.} \qquad (30).$$

When the activation of one of the muscles that takes care of the moment around the head of the femur(e.g. the M.Psoas) increases above normal, the force F_{opt} also increases. From formula (30), in which p and H retain the same value, it follows that e must also increase. This means that the weight of the upper part of the body must be displaced further backwards. This can be done by turning-over the pelvis backwards so that the sacrum will stand steeper. This is not possible, unless the M.Psoas is mad longer. A second procedure is to have the lordosis increased. People with a contraction or shortening of the muscles, that take care of a clockwise moment, will continuously lordosize to a greater extent their lumbar spine. When the shortening of the muscles increases, the moment of the hip also xxxxxxxxx increases, the lumbar spine lordosizes more, until it is no longer possible to produce an equilibrium only by one's own weight; the muscular groups that provide an anticlockwise moment of the hip also need to be tightened.

If the muscle that has been shortened runs from femur to lumbar spine(like the M. Psoas),the pelvis will turn over forward in the first instance and a substantial tensile force will be exerted on the lumbar spine.
Good posture implies that the line of gravity lies behind the head of the femur. In this case the principal postural muscles like the M.Psoas and the M.Rectus femoris are slightly tightened and take care of the equilibrium around the hip-joint. Should these muscles have been shortened or be hypertonic,then the equilibrium is disturbed. Mostly an increased lordosis is the consequence of it,and also a tightening of the phasic glutaeal muscles,which lie on the other side of the head of the femur. In its turn,this results in low back complaints,and tiredness that occurs sooner when standing.

6.3 Postural-mechanics of pregnant women.

In order to study the postural control system(as described above) with application to some real life problems,a research was made into the posture of women who were in the last month of pregnancy,and into the alteration of this posture when two weeks after partus had elapsed. To this effect the contours,the stabilograms,the weights and the lengths of seven women were measured before and after child birth.
In order to be able to observe high percentage changes in body-weight,women having a normal weight before pregnancy were studied. The first remarkable result of this research was that after partus all women were about 1-2 cms. shorter in height than before partus. The cause of it was the larger curvature of the spine after the partus. Continually,after partus,a thoracic and a lumbar increase of the curvature of the spine(indicated by points A & C in fig. 41) were observed. This is contrary to the generally accepted opinion.
From the contour-picture this curvature was calculated mathematically.
Thoracic and lumbar curvature means here the curvature of the most dorsal and the most ventral part of the kyphosis and the lordosis respectively.
The increases in curvature(measured after childbirth) of the seven women are shown in the table below.

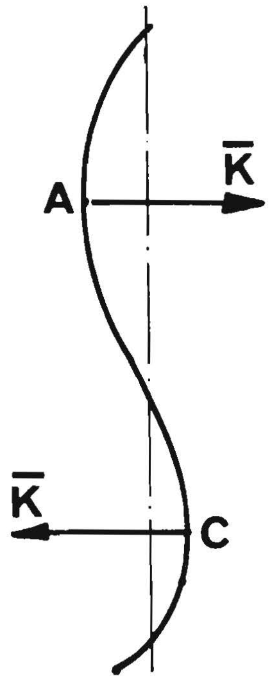


fig. 41. The increase in curvature after childbirth was measured at the points A and C,

no.	thoracic $\Delta K (mm^{-1})$	lumbar $\Delta K (mm^{-1})$
1	$6{,}4 \cdot 10^{-4}$	$20{,}0 \cdot 10^{-4}$
2	5,3 . "	15,9 . "
3	5,5 . "	16,5 . "
4	4,2 . "	8,8 . "
5	7,3 . "	20,2 . "
6	7,1 . "	14,0 . "
7	11,3 . "	22,6 . "

Table 1. Increase in the curvatures(Δ K) after childbirth.

This increase in curvature can easily be explained in terms of the conditions of equilibrium around the head of the femur. In fig. 42 once againthe kinetics state of this equilibrium condition is schematized. Therein,H represents the head of the femur and B the pelvis.

The pelvis balances on the head of the hip and the two principal forces affecting the pelvis are F and P,respectively the weight of the body above the head of the femur and the muscular force P.

The force F lies behind the head of the femur(see chapter 5),and in order to obtain equilibrium around this head,the muscular groups on the ventral side must tighten. In this respect the most important muscular group is the M.Psoas.

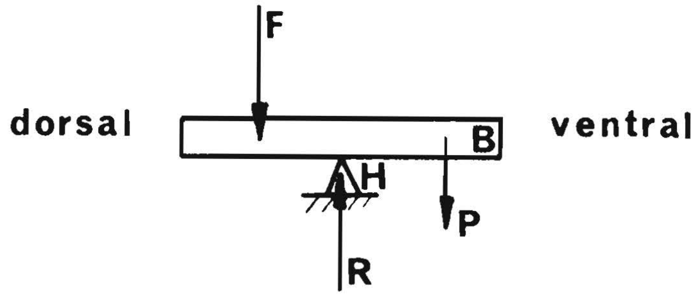


fig. 42. Equilibrium of the pelvis B on the head(H) of the femur. Schematic drawing.

The above situation exists in most cases. When,however,an extra-weight,G,is added on the right side of the head of the femur,e.g. in case of pregnancy,fig.42,shows that there are two possibilities to effect a new equilibrium.

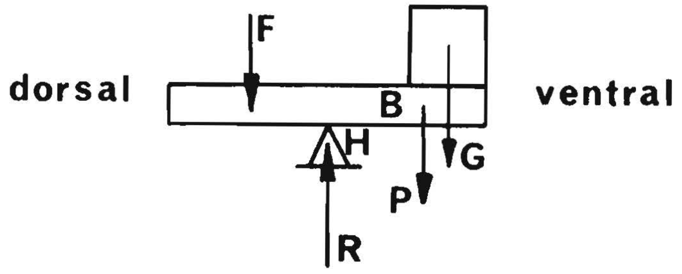


fig. 43. In pregnancy,P must decrease as a consequence of effecting an equilibrium with the help of the extra weight in the abdomen.

The first possibility is displacing the weight(F) of the trunk,head and arms further backwards. This results in an increase of the anticlockwise moment about H in proportion to the clockwise moment due to the added force G. Consequently,extra muscles must be employed to take the line of action F further backwards.

The second possibility in order to effect equilibrium entailing the relaxation of the M.Psoas is more plausible,because the M.Psoas had already been tightened. As a consequence in total less energy is employed and the total reaction force (R) on the head of the femur does not increase. On relaxing the M.Psoas such that the decrease of the clockwise moment due to a decrease in P is as large as the increase of the anticlockwise moment due to the added force G, equilibrium will exist again. Thus, when standing unconstrainedly during pregnancy, the tightening of the M.Psoas will be less. This relaxation also has an immediate consequence as to the lumbar spine, the M.Psoas having its origin here. Refer fig. 44, wherein A is the clasp-point of the spine in the pelvis, R is the resulting force due to body weight and muscles lying behind A plus the muscles lying between the spine and the pelvis (such as the dorsal extensors and the M.Quadratus Lumborum), P is the force due to the M.Psoas (causing a clockwise moment).

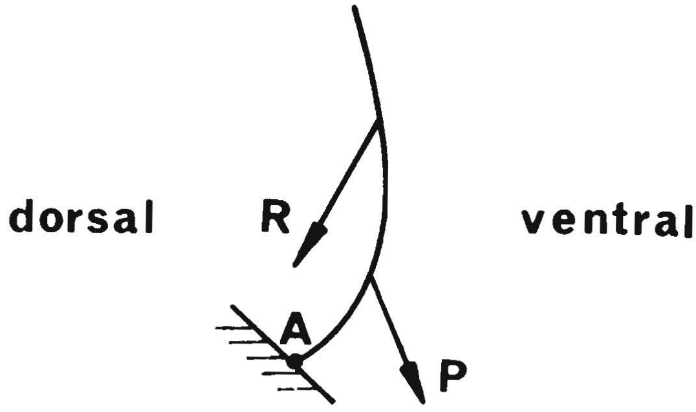


fig. 44. Clasping of the lumbar spine in the pelvis A with the help of the affecting forces P and R.

In order to have equilibrium with P, the force R adapts so that its anticlockwise moment about A will balance the clockwise moment of P about A.
Consequently the sacrum-iliacum joint (clasp-point A) will be loaded with bending as little as possible. Thus, when the M.Psoas as a result of pregnancy tightens less, the whole lumbar spine will be loaded less in bending. This change in bending moment will take place in a linear relationship to the change of weight in the abdomen G, because the muscle sites do not vary. Consequently every change in moment will only depend on the changes in force, so that we can write

change in moment, $\Delta M \propto \Delta P$, the change in force P.

However, according to fig. 43 $\Delta P \propto \Delta G$, due to increased abdominal weight.

Hence, $\Delta M \propto \Delta G$ (31).

The curvature is determined by the following formula:

$$K = \frac{M}{E.I}$$

in which K represents the curvature

M represents the bending moment, and
E.I " " rigidity of the lumbar spine.

This formula can also be written

$$\Delta K = \frac{\Delta M}{E.I} \qquad (32)$$

in which ΔK and ΔM represent the changes in curvature and in moment respectively.

For adults: E = constant and $I \propto d^4$.

Hence $E.I \propto d^4$.

Fig. 29 implies that $G \propto d^4$

so that $E.I \propto G.$ (33)

(this equation can only be used for adults, because for children during growth: $E \propto d^2$, see chapter 5).

Equation (32) can now be written as

$$\Delta K \propto \frac{\Delta M}{G} \qquad (34)$$

Applying equation(31) this becomes

$$\Delta K \propto \frac{\Delta G}{G}$$

or $\Delta K \cdot G \propto \Delta G$ (35).

From this formula and also from the above it appears that the product of lumbar change in curvature and weight is proportional to the increase in weight in the abdomen, when the equilibrium around the head of the femur is determined by the M.Psoas.

For the seven women examined, $\Delta K.G$ and ΔG have been calculated and put together in the table that follows. In fig. 45 this relation is represented graphically.

In accordance with formula 35, which is based upon hypotheses, through the seven points of measurement a regression has been drawn that is both linear and passes through zero-point.

Passing through zero-point is obvious because in case of no change in weight, no change in curvature will occur.

In fig. 45 this relation is represented graphically, the regression fits well.

No.	G(N)	ΔK(mm^{-1})	G. ΔK ($N\ mm^{-1}$)	G(N)
1	704	20.0 . 10^{-4}	1408.10^{-3}	87
2	587	15.9 . "	933. "	72
3	629	16.5 . "	1036. "	64
4	622	8.8 . "	547. "	56
5	714	20.2 . "	1440. "	91
6	704	14.0 . "	984. "	49
7	603	22.6 . "	1360. "	146

Table 2. The body-weight G, the change in curvature ΔK, the product (G. ΔK) and the change in weight ΔG of the seven women studied.

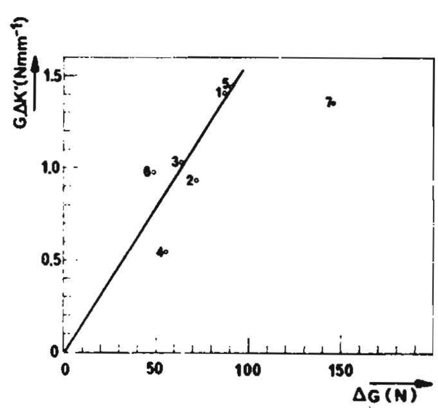


fig. 45. The product G. ΔK plotted against the change in body-weight.

No. 7 is an exception and does not conform to the rectilinear relation of formula(35); herein another mechanism must contribute to the equilibrium around the head of the femur.

With respect to no. 7 the first thing that strikes is the large ΔG(twins).

When studying the contour pictures of this woman, taken before and after childbirth(fig. 46), it is to be seen that in the lumbar area the curvature after childbirth has flattened much. Also, noting the lumbar region, it seems that

during pregnancy the woman could not decrease her lumbar lordosis anymore and with the respect to Δ K, consequently, reached an upper limit (the order of magnitude is the same as in the cases 1 and 5).
The explanation for this is that,according to fig. 44 the M.Psoas is entirely relaxed. Yet,in order to have equilibrium with the very large weight in the abdomen,this muscle cannot relax further; consequently those muscles will have to be tightened that provide a counter-clockwise moment around the hip. The most important muscle of this muscular group is the M.Glutaeus Maximus. This muscle runs from the upper side of the hip to dorsal side of the pelvis and is not attached to the lumbar spine. **As a result the tightening of the Glutaeus has no direct bearing on the curvature K and on K of the lumbar spine. Hence this case does not fall on the linear regression line of fig. 45.**

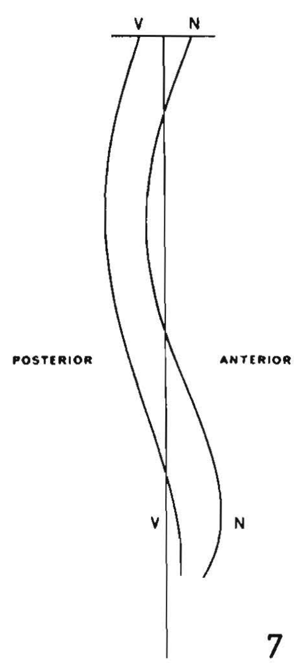


fig. 46. Contour pictures of the twins carrying woman in standing posture. V : during pregnancy and N : after partus.

When summarizing it can be said that as a consequence of pregnancy,in proportion as the weight in the abdominal cavity increases,the lumbar spine will be less curved This results in an increase in length of 1 to 2 cm. and in a flat back.
The above may imply that,should a woman's lumbar lordosis not decrease,shortened muscles are the cause for it,particularly those of the M.Psoas.
In this case a woman will find difficulty in arranging favourably equilibrium around the pelvis,resulting in an excessive load on the sacrum-iliacum joint.

6.4 Mechanics of sitting posture; implications to comfortable chair design.

The form taken by the spine when sitting with only one support at the level of the shoulders,can be compared with a spring-column that is hinged at the base and,at the top,is clamped to a support that can move freely in a vertical direction(fig. 47). This column has two stable positions viz. one on the left and one on the right side,which can be retained without external forces.

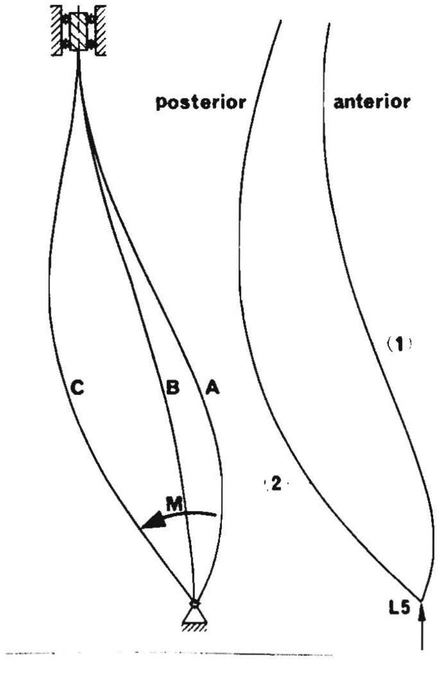


fig.47. Click-clack phenomenon. The spine (right) has two stable positions, just like the spring-column (left). By a momentarily exerting moment in the bottommost point, the form can 'click' from one stable position into the other.

In order to put the column from one of these stable positions into another,only for a short while an auxiliary moment at the bottom hinge needs to be momentarily appli

In order to confirm this,we have taken two contour-pictures of a very flexible girl (aged 22) in sitting posture(fig.47). These pictures show clearly the bi-stable behaviour,or,in other words,the so-called click-clack phenomenon.

Transition from sitting posture into lying posture,vice versa.

For a lasting equal sitting posture,the stable position 1 in fig.47 must be preferre (circulation,no strained ligaments as in position 2). During transition from sitting posture into lying posture this stable position must be maintained as long as possible. Among other things this is important in connection with the reclining chair. In fig. 48,G is the total weight of the trunk,the head and the arms.

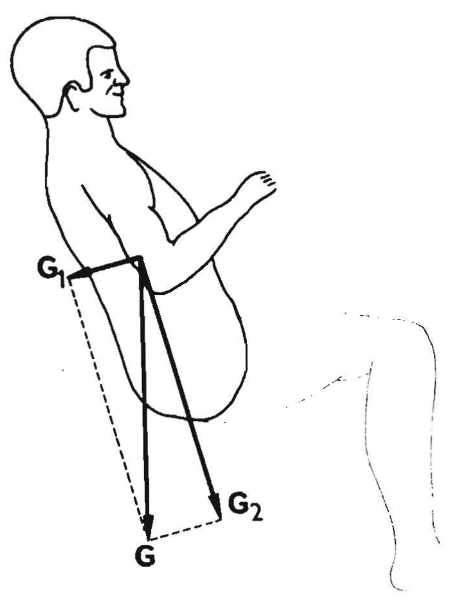


fig. 48. The components G_1 and G_2 of G. G is the total weight of trunk,arms and head. G_1 and G_2 are respectively perpendicular and parallel to the back of the chair.

This force G can be divided into two forces,G_1 and G_2,which are **acting** respectively perpendicular and parallel to the back of the chair.

When tilting backwards the back of the chair,the component G_1 will increase.

This component G_1 procures the moment M of fig. 47,which is the cause of the transition from stable position 1 into stable position 2(click-clack). In order to maintain stable position 1,the need exists to introduce an extra force F on the back, which must have the opposite direction of force G_1(fig.49). This force F must act on the upper edge of the pelvis.

There are now two points of support at the back,viz. one the height of the shoulders and the other one at the upper edge of the pelvis. These points of support do exist if an even plane,the tangent plane,is placed behind and against the back,which has adapted the stable position 1(line a - a in fig. 49).

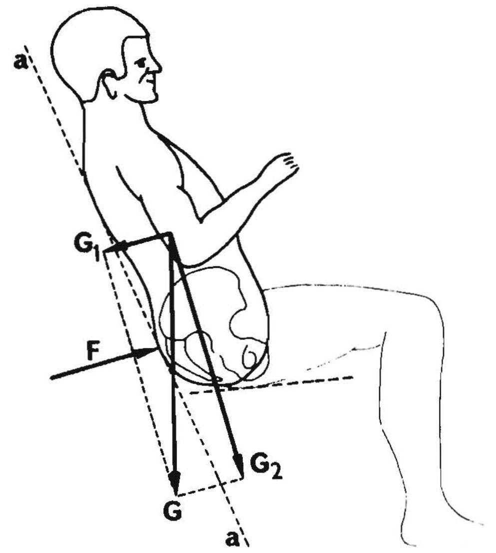


fig. 49.F,an extra force,looks after the equilibrium of moment,so that the stable position 1 is maintained.

In practice this plane is represented by the back of a chair.
Both points of support must be maintained during transition into lying posture, and for this reason the back of the chair must follow and conduct the back and the pelvis. This means that the mechanical rotation-axis of the back of the chair must coincide with the required biomechanical rotation-axis between the back and the upper legs.
In general,in order to maintain a good support,the condition should be fulfilled that the mechanical rotation-axis between two planes of support ought to coincide with the required biomechanical rotation-axis between the corresponding rotating parts of the body.

In connection with the transition from sitting posture into lying posture,vice vers this means that

- the rotation-axis between the back of the chair and the seat should coincide with the biomechanical rotation-axis between the trunk and the upper legs
- also the rotation-axis between the legs-rest and the seat should coincide with the biomechanical rotation-axis between the lower legs and the upper legs.

The situations of the required biomechanical axes have been determined experimentally*). Fig. 50 shows a sketch of the measuring equipment and the geometrical parameters.

*) Bougie,T.H.M. and Meeuwissen,L.M.K. (1972): De Rolstoel, Report University of Technology, Eindhoven, Holland.

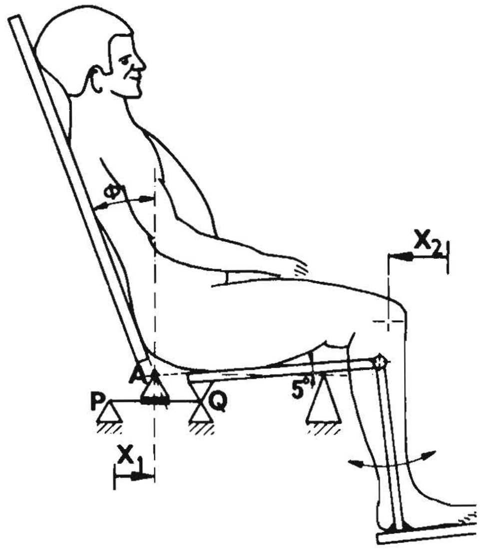


fig. 50. A schematic sketch of the measuring equipment, containing the geometrical parameters.

The back of the chair and the seat have the shape of even planes upholstered with foam-rubber(thickness 15 mm). The edge of the plane of the back of the chair lies 10 cm. above the seat, so that a possibility for sagging of the soft parts(buttocks) exists.
The back of the chair can rotate around a point A that lies the same height as the seat and can produce an angle Φ $(0^{o} \leqslant \Phi \leqslant 90^{o})$; it can also translate over a distance x_1 along a horizontal axis P - Q.
By adjusting Φ and x_1, irrespective of each other, the back of the chair can be placed in any desired position.
The knee is considered a possibility for rotation of the tibia with respect to the femur around one certain axis.
In reality, however, the tibia rolls over the femur, which implies a small displacement of the rotation-axis. Henceforth this displacement is neglected in the following text-lines.
Whilst the back of the chair is tilted, the knee displaces.
This displacement which equals the displacement of the femur-tibia rotation-axis is expressed by the co-ordinate x_2 in fig. 50.
The feet are supported by an even plane, in such a way that the upper legs are dis-

tinctly touching the seat.
In order to effect an unconstrained sitting posture, the Φ -values being small, the seat has been mounted having an angle of five degrees.
As to great Φ -values this angle does not exert any influence, hence it has been maintained during the whole measuring procedure.
Measurements were done as follows.

Starting from the initial position Φ = 0 degrees, the angle of the back of the chair was changed phasically by 4 - 6 degrees: $\Phi = 0^o \rightarrow 90^o \rightarrow 0^o$.
After each Φ -phase, x_1 was adjusted such until the desired support of the pelvis existed, and the patient said that hewas sitting comfortably. After this, x_2 was measured.
Starting from $\Phi = 60^o$ the pelvic position was affected by the bi-articular muscle around the pelvis of which the M.Psoas and the M.Rectus Femoris are the most important. **As the lying posture (with stretched legs) became the final position, both lower legs were placed at an angle of 45^o to the vertical, starting from $\Phi = 50^o$.**
Starting from $\Phi = 90^o$, in phases of 4 - 6 degrees, measurements were repeated in the opposite direction.
Fig. 51 shows graphically the measuring results.

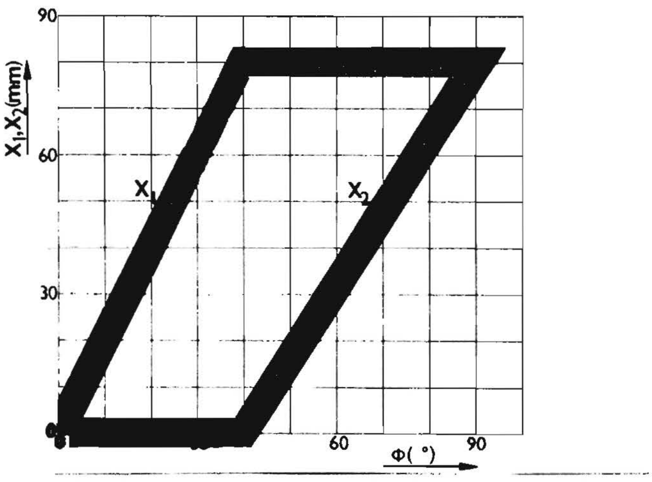


fig. 51. x_1 and x_2 as a function of Φ .
Number of measurements: 12.
$0^\circ \leqslant \Phi \leqslant 40^o$: x_1 is a linear function of Φ ,
x_2 is a constant.
$40^o < \Phi \leqslant 90^o$: x_2 is a linear function of Φ ,
x_1 is a constant.

We note that no distinct correlation has been observed between build or sex,and x_1, x_2.
In case of $30^\circ \leqslant \Phi \leqslant 50^\circ$ it appeared to be difficult to find an unconstrained, comfortable sitting posture(patient's subjective remark).
An explanation of the monitored results must take into account a sliding down and slipping of the pelvis combined with a bending of the spine.
In fig. 52 the pelvis has been drawn in the case of $\Phi = 0^\circ$(line 1)

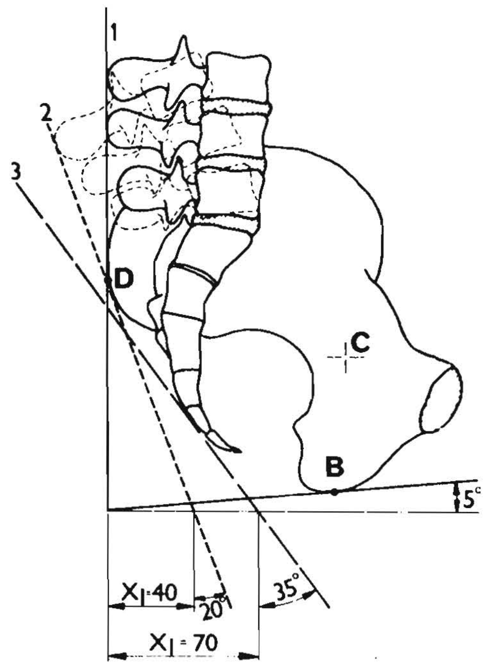


fig. 52. The pelvis with some lumbar vertebrae in case of $\Phi = 0^\circ$ (line 1).
B: the point of contact of the pelvis and the seat.
C: the hip-joint(fixed point in pelvis).
D: the point of contact of the pelvis and the back of the chair in case of $\Phi = 0^\circ$.

Point C is the hip-joint(a fixed point in the pelvis).
Points B and D respectively are the points of contact with the seat and the back of the chair in case of $\Phi = 0^\circ$.
Point C is connected with the knee through the femur.
Measurements(fig.51) gave the result that in case of $0^\circ \leqslant \Phi \leqslant 40^\circ$,$x_2$ is constant; consequently the hip-joint then does not remove. We can draw the back of the chair having a number of Φ -values,moving along a distance x_1,and depending on Φ according to fig.51.
Fig.52 shows these situations in case of $\Phi = 20^\circ$(line 2) and in case of $\Phi = 35^\circ$ (line 3).

It appears that in case of $0^\circ \leqslant \Phi \leqslant 25^\circ$ the lines defining $x_1(\Phi)$ in fig.51, will touch the pelvis at point D(fig.52). Consequently two points of the pelvis exist, viz. C and D, that do not move. This means that the pelvis remains in the same position and consequently does not rotate or translate in the area $0^\circ \leqslant \Phi \leqslant 25^\circ$

The rotation of the back of the chair is followed by the rotation of the lowermost lumbar vertebrae. In fig. 53 these vertebrae have been drawn for the case of Φ being 20°.

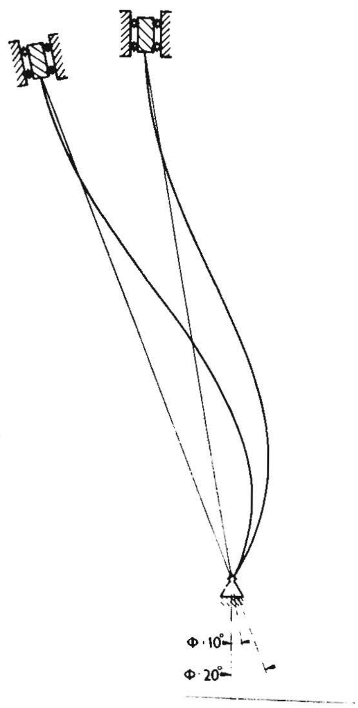


fig. 53. **The spine, according to fig. 47, drawn for $\Phi = 10^\circ$ and $\Phi = 20^\circ$.**

After reaching the maximum rotation of the lowermost vertebrae(approx. 14° rotation with respect to each other), the back of the chair does not touch the pelvis in the situation of figure 52(line 3). Then the pelvis rotates around the hip-axis C in the case of $25^\circ < \Phi \leqslant 40^\circ$ until it touches the back of the chair again after each step of Φ. This appears from the measuring results: x_1 increases, x_2 remains constant.

In this stretch the seat-bones slide over the seat. The frictional forces and the less stable position of the pelvis, are, as was noted before($30^\circ \leqslant \Phi \leqslant 50^\circ$) the cause of the difficult sitting posture.

This rotation of the pelvis continues until the sacrum-coccyx touches the seat(fig. 54) at point E. This occurs in case of $\Phi = 40^\circ$.

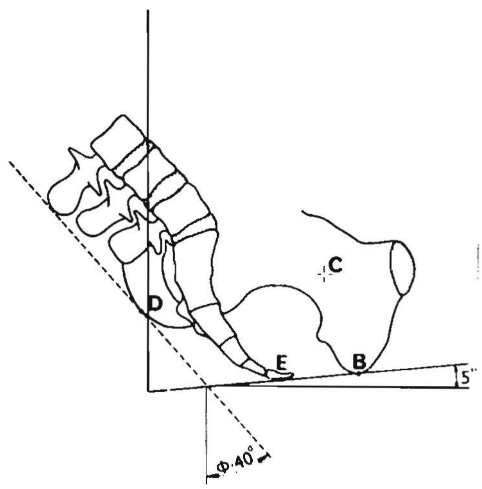


fig. 54. Starting from $\Phi = 25^{\circ}$, the pelvis topples over until the sacrum/coccyx touches the seat at point E ($\Phi = 40^{\circ}$).

From this position the pelvis rolls over the sacrum($40^{\circ} < \Phi \leqslant 90^{\circ}$) to a stable position in which the edge of the pelvis and the lumbar vertebrae are supported by the back of the chair and the seat($\Phi = 90^{\circ}$).
During this last phase the hip-axis C moves backwards(fig. 51: $x_2 = 0 \rightarrow 75$ mm.).
Initiating from pure rolling this shift in a horizontal direction is 80 mm.
Notwithstanding negligence of the occurring slip, this corresponds well with the measuring(75 mm.).

<u>Summary.</u>

The desired support for the back as well as the biomechanical rotation-axis of the trunk with respect to the upper legs, have how been determined. The site of this rotation-axis depends on Φ :

(fig.55):	$0^{\circ} \leqslant \Phi \leqslant 25^{\circ}$	D(upper edge pelvis)
	$25^{\circ} < \Phi \leqslant 40^{\circ}$	C(hip joint)
	$40^{\circ} < \Phi \leqslant 90^{\circ}$	according to dotted line from E to F.

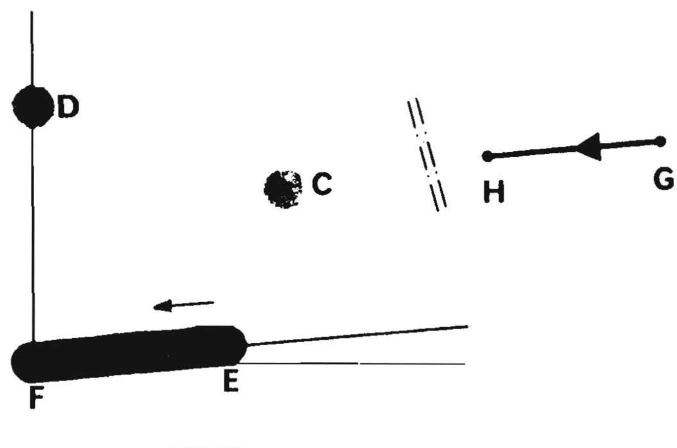


fig. 55. The site of the rotation-axis depends on the position of the back of the chair(see text).

The biomechanical knee-axis moves in case of $40^\circ \leqslant \Phi \leqslant 90^\circ$ linear with Φ from G to H(75 mm.). In case of $\Phi = 0^\circ$, G is here the spot of the knee.

Conclusions; directives for the design of a reclining chair.

1.For taking up a good sitting posture,the support of the pelvis is a primary demand.
2. Concerning a good support whilst tilting ,the back of the chair should rotate around an axis like the one in fig.54.
3.In reality,the keen transitions the above model comprises,do not exist(means of measuring,simplification of measurement values). A relatively short transition stretch may be observed between two sites of the rotation-axis.
4.It is necessary that the back of the chair's lowest edge lies approx. 10 cm. above the seat(fig. 52,line 2).
5.The area being the consequence of $30^\circ \leqslant \Phi \leqslant 50^\circ$,should be avoided because the position of the pelvis is unstable then.
6.Should an adjustable leg support have been mounted,this support ought to rotate

around an axis running through the knee-joint. This axis should translate parallel to the knee-joint(fig.54: G ⟶ H).

7. Two surgical operational applications.

The mechanical and mathematical analyses of the spine(presented earlier) provide the foundations for the development of new methods of treatment of the pathological spine. Herein we will discuss the following two operative methods:

1. the correction of spondylolisthesis
2. the correction of scoliosis.

7.1 Spondylolisthesis.

Spondylolisthesis occurs with approximately 2 - 4% of the people of the developed countries. As a consequence of a defect(spondylolysis) in the interarticular part of the neural arch,the vertebral body together with the above lying spine has slipped forward(olisthesis). Besides irritation of the cauda and the nerve roots a particularly bad posture can result,which in the long run can give rise to several complaints(fig.56).

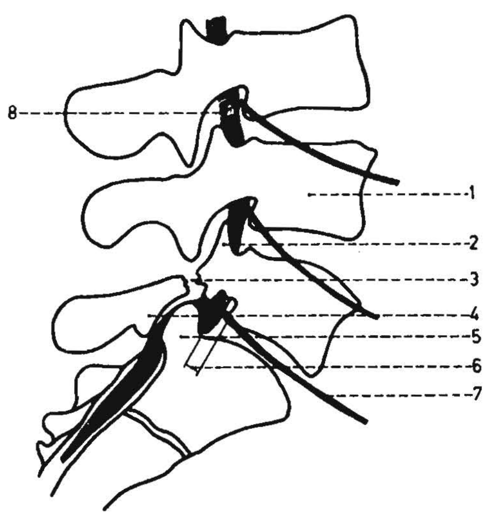


fig.56. As a result of a fracture in the neural arch, the vertebral body slips to the ventral side. 1.Vertebral body,2.Superior articular process of L5, 3.Lysis, 4.Neural arch, 5. Superior articular process of S1, 6.Olisthesis, 7.Neural root, 8.Cauda (spinal marrow).

The result of every therapy applied so far is only chiefly temporary or partial abolition of the pain,without normalizing the mechanical situation and improving the posture. For normalization of the mechanical situation(in cases of spondylolisthesis) first of all the olisthetic vertebra must be placed again on its original spot. The necessary force(calculated for this purpose) is introduced with the help of stainless steel wires that are attached to the processus spinosi of the two vertebrae lying above the olisthetic vertebra(fig. 57).

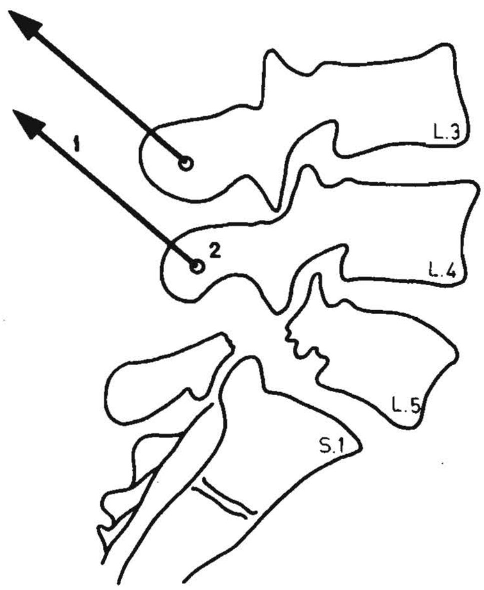
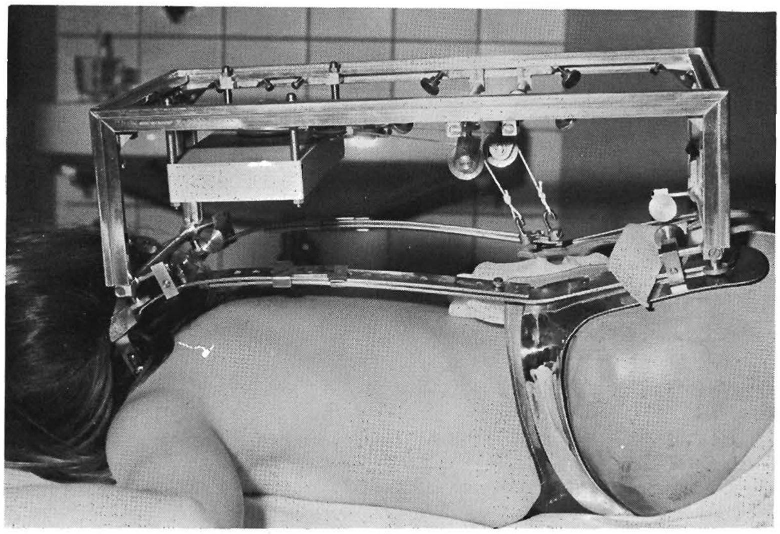


fig. 57. **The traction is attached to the processus spinosi of L3 and L4.**

The wires are put through little holes made in the processus spinosi with the aid of a pair of punching-tongs. The isotonic tensile force is applied to the wires by means of a springmotor that has been hung in a frame(fig. 58). When changing the wire-pulley on the shaft driven by the spring-motor,another force can easily be procured

The direction of the force,exerted on the olisthetic vertebra,is adjusted by means of conducting wheels for the wires that can be transposed in respect of the frame. A winding-mechanism for the wires and a system of pulleys enables both steel wires to be made tense gradually and at the same time. The frame is positioned on the body by means of a Milwaukee Brace.

The magnitude of the force needed for the repositioning is calculated with the help of a biomechanical shear model,to which effect measuring of the grade of olisthesis,the disc diameter and height,is done in a lateral X-ray picture.



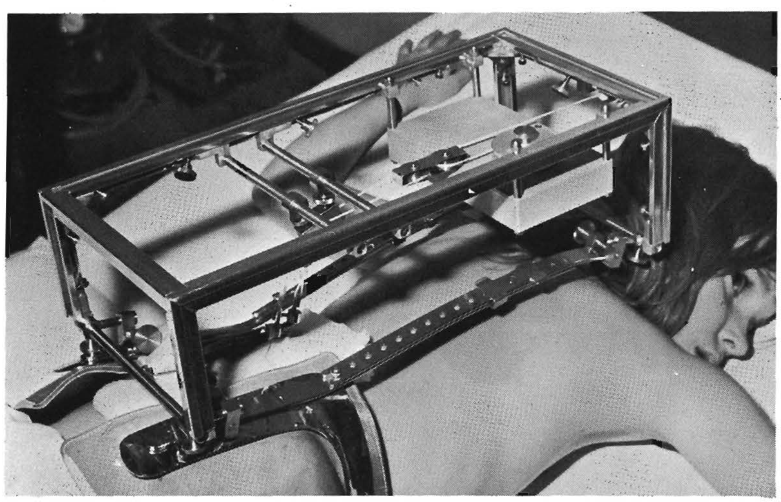


fig.58. Traction-apparatus for the repositioning of spondylolisthesis. Isotonic forces are exerted by means of a spring motor,directly on the spine by means of wires passing through the skin.

At the level of the olisthesis(which,in 80% of the cases,means the level of the fifth lumbar vertebra-first sacral vertebra) the disc-diameter,the disc-height and the extent of shear,are measured and corrected by the X-ray projection factor. The shear is measured with respect to the vertebra lying beneath and expressed in percentages of the diameter of this vertebra(fig.59).

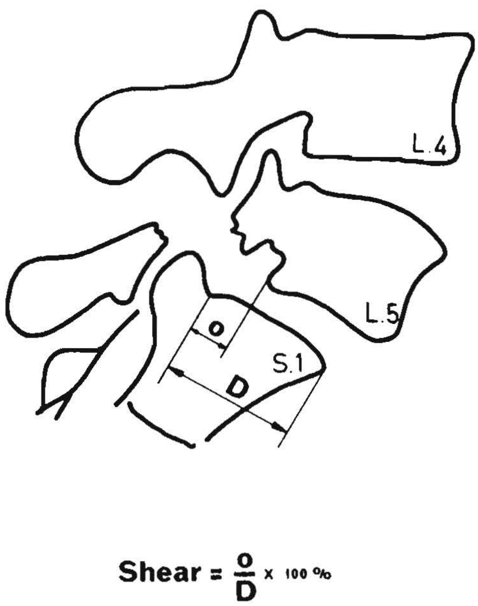


fig. 59. **The shear is measured with respect to the vertebra lying beneath and expressed in percentages of the diameter of this vertebra.**

The applied procedure distinguishes between the following cases:

(i) olisthesis smaller than approx. 30%, (ii) olisthesis larger than approx. 30%.

<u>If the olisthesis is smaller than approx. 30%,</u> then relatively small forces (magnitude approx. 40 N.) on the processus spinosi of the two vertebrae lying above, are adequate to stretch the tissue of the annulus fibrosis and of the surrounding ligaments to reposition the olisthetic vertebra completely.
In the case of a spondylolisthesis on the level L5 - S1, the discs L4 - L5 and L3 - L4 transmit the correction-force through to the olisthetic level. The correction process takes approx. 36 hours and occurs with a fully conscious patient.
In order to keep the repositioning during operation, when the wires are introduced, an intercorporal spondylodesis is applied from dorsal (fig. 60).

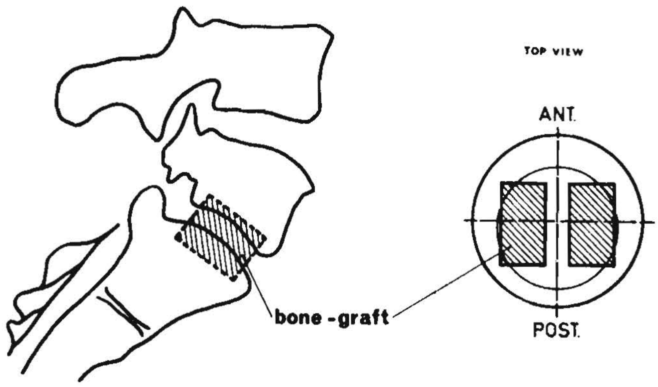


fig. 60. **Lateral and top view of the position of the grafts in the intervertebral disc.**

To this effect two bone-grafts from the crista are put into intervertebral space. along both sides of the dura. In order to provide compatible growing together with the vertebral bodies, grooves are milled in both boundary plates of the intervertebral space. During the time of growing together of the bone-grafts the traction is maintained through the frame of fig. 58.

If the olisthesis is larger than approx. 30%, the repositioning force needed becomes too big to be sustained by the allowable fracture strength of the spinous processi. In such cases, the larger part of the annulus fibrosis and the ligaments on the olisthetic level are cut and the traction required for a complete correction need not be larger than the lumbar spine's own weight.

In these cases it is practical to prepare the vertebral bodies in such a way that these can grow together immediately(by milling off the cartilage layers of S1 and L5), so that no bone-grafts need be introduced for fixation. During the post-operativ phase, the patient is revalidated in order to obtain a non-pathological posture of the body. In this connection especially the M.Psoas plays an important part(fig.61).

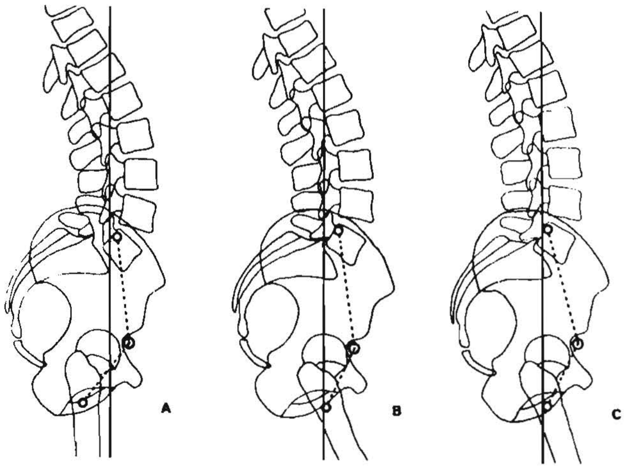


fig. 61. The lumbo-sacral region in case of Spondylolisthesis. The dotted line gives the position of the M.Psoas. A: situation before operation. B: the state directly after operation. C: the state after some strechting of the M.Psoas.

In most cases this muscle must be lengthened by exercises for that purpose;this is also necessary with respect to changed geometrical proportions that are the consequence of the repositioning.

Fig. 62 shows a case of repositioning and fixation

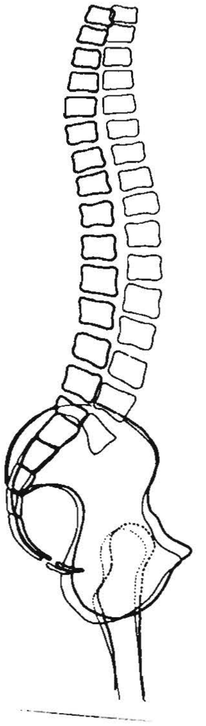


fig.62. A case of spondylolisthesis greater than 30%,before and after repositioning and fixation.

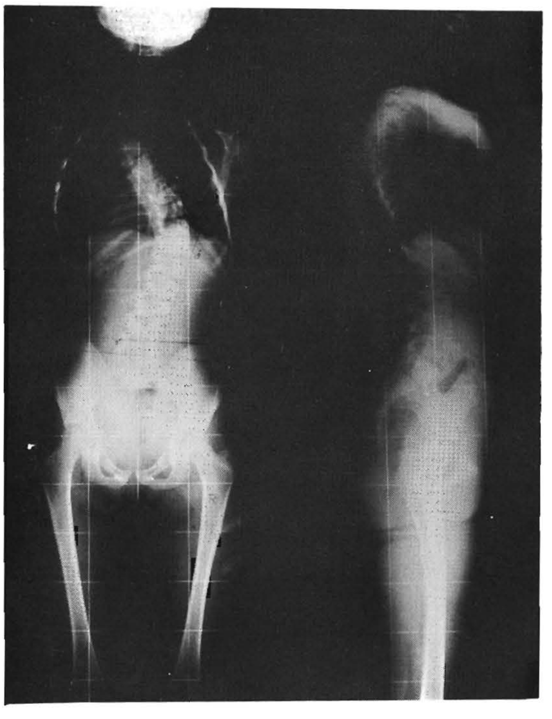


fig.63. X-rays of a patient with scoliosis

During all operations performed, in addition to the redressing of complaints, the excessive lordosis (present, if any), the steep sacrum base incline and the psoas-contracture were also corrected.

7.2 Scoliosis.

Scoliosis is a lateral curvature of the spine attended by an axial rotation of the vertebrae (fig. 63). For correction of scoliosis (caused by a non-structural deformation of the vertebrae), a method is developed to correct this deformation, provide an erect posture and thereafter maintain the corrected form without hindrance to the growth. The principle of the correction method involves correction of scoliosis with the help of transverse forces shown in figure 64.

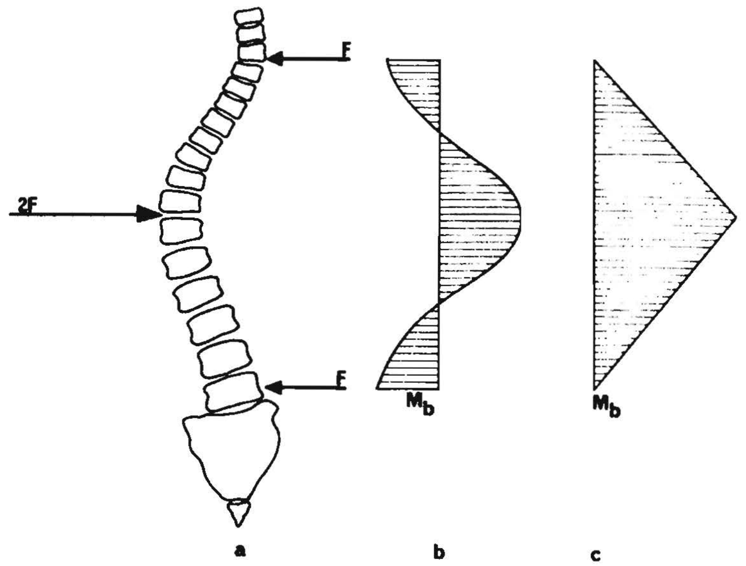


fig. 64. a. Transverse forces needed to correct a scoliosis.
b. Graphic representation of the bending moment variation needed to correct a scoliosis.
c. The bending moment caused by the transverse forces.

Figure 53 shows (i) the bending moment distribution, in the spine of a certain scoliosis patient, that is needed to correct the scoliosis completely and (ii) the bending moment introduced by corrective transverse forces on the spine at three levels. The bending moments correspond well; therefore these three transverse forces are sufficient to correct the scoliosis. The forces are exerted direct on the three vertebrae by means of bars (fig. 65).

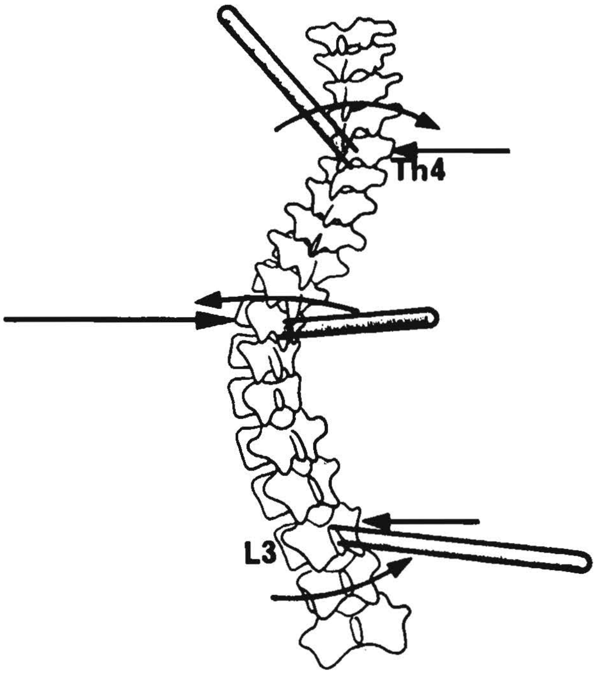


fig. 65. **Spatial view of a scoliosis with the three bars to correct the curve.**

In the thoracic region these bars are attached to the processus transversi by means of a clamping-device. In the lumbar region attachment is made to the neural arch and the processus spinosus. The forces are introduced from outside the skin. (fig.66) Two forces are introduced into each bar, because the force causing the transverse force (F_1) yields a moment anticlockwise with respect to B(the geometrical centre of the body), whereas the rotation-correction demands a clockwise moment. In order to generate these six forces(two for each vertebra), an external frame of two extra rods is designed to connect the uppermost and lowermost vertebral bar. Between these rods and the middlemost vertebral bar are two screw-thread connections(fig. 67).

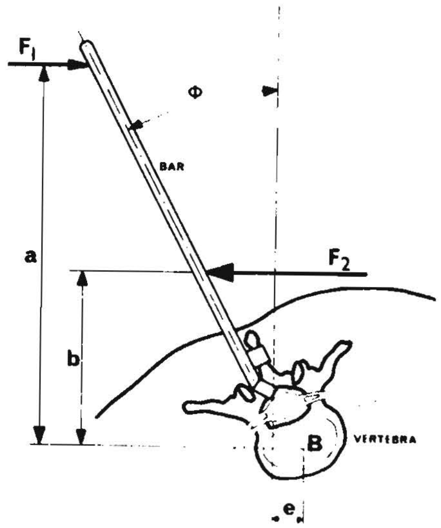


fig.66. Two forces are needed to bring the vertebra back to its initial position.

$F_2 - F_1$ causes the transverse force.

M, the torsional moment around B, is $F_1 \cdot a - F_2 \cdot b$

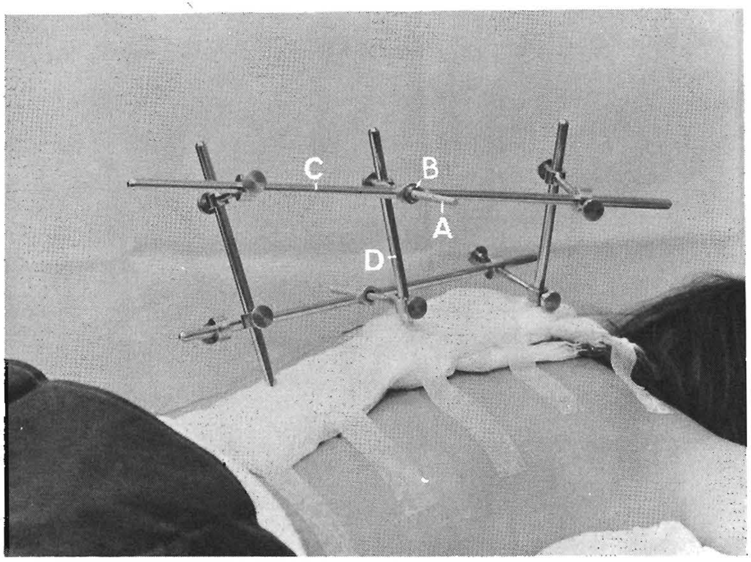


fig.67. Patient with scoliosis corrector. The function of the external frame is to place the vertebrae "in line".

A : screw-thread

B : correction nut

C : connection rod

D : bar connected with the vertebra.

By means of gradual turning-on of the nuts on the screw-threads, the middlemost vertebra can be placed fully in line with the other two.
The turning-on of the correction nuts takes place with the help of a moment producin device, from which by means of a calibrated scale, the transverse force introduced ca be read directly. When the correction force reaches a value of about 50 N., the correcting is stopped, and as to the next phase of correction is waited until the transverse force will have dropped to half of the said 50 N., due to relaxation of the spine and the thorax. After that the 'turning-on' of the nuts is started again.
The correction process takes place very gradually. With the patient of fig.68 havin a scoliosis of 46^{o}, it took one week for complete correction. The process occurs wit full knowledge of the patient. The patient does not find any inconveniency from it.

When the spine has been over-corrected to approx. 5°, the fixation is affected. This fixation must take place in order to prevent the spine taking up again the scoliotic form. The staightened spine is fixed with the help of the existing Harrington-apparatus(fig. 68).

The above method for the correction of scoliosis is not yet an optimum one. Severe cases are giving problems and the treatment consists of two operations,viz. 1) implantation of corrector and 2) implantation of fixator.

A change in treatment having the consequence that in future only one operation is needed in order to come up to both requirements,is an ambition to be realized. Research in this field into the most optimum technical realization of it is being made.

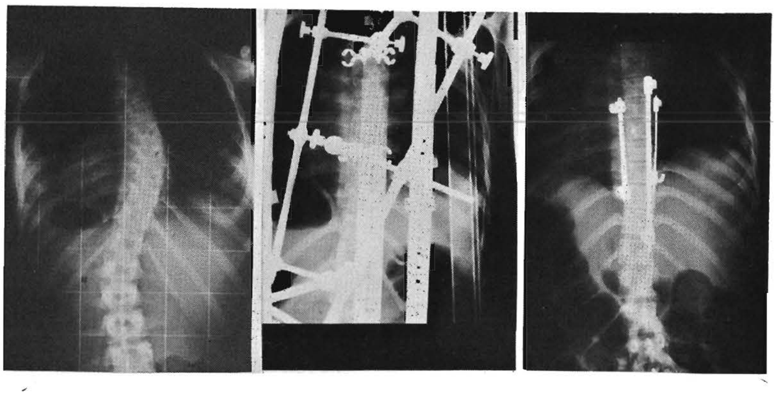


fig. 68. Patient of 16 years with a scoliosis of 46°. Left X-ray is before operation. Middle is during the correction period of one week. Right is after implanting the Harrington-equipment to fixate the spine.