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THE EFFECTS OF KNEE BRACE HINGE DESIGN AND PLACEMENT ON JOINT MECHANICS

P. S. WALKER*, J. S. ROVICK and D. D. ROBERTSON

Clinical Bioengineering, Veterans Administration Medical Centre, West Roxbury, MA, U.S.A. and Orthopedic Biomechanics, Brigham and Women's Hospital, Boston, MA, U.S.A. (Teaching Affiliates of Harvard Medical School)

Abstract—A computer model of 23 knees was obtained by embedding, slicing and digitizing the bone outlines and ligament co-ordinates. Using co-ordinate transformations, various three-dimensional motions were imposed on the knees, and calculations made of femoral–tibial contact error, contact point locations and ligament lengths. Significant deviations in these parameters were noted for abnormal motions including the elimination of internal–external rotation and a–p displacement and the misplacement of a hinge producing correct motion. The resulting mismatch could result in shear in soft tissues, cuff-to-skin slippage and inaccurate ligament length patterns.

INTRODUCTION

Knee braces are used for a variety of applications including the prevention of injury, stabilization of chronic instability, and protection after injury or surgical repair (Butler *et al.*, 1983). Braces have recently been classified as prophylactic, rehabilitation, or functional (A.A.O.S., 1985). For application to normal healthy knees or to the chronically unstable knee, the motion provided by the hinges may be relatively less important than the ability of the brace to limit the displacements and rotations between the femur and the tibia. Any mismatch in motion between the hinges and the joint itself will be taken up by soft tissue deformation and by cuff-to-skin slippage. For rehabilitation however, there are more stringent requirements on the motion which the external hinges produce, in order to prevent further injury and to allow soft tissue healing to occur within correct length limits. Except for short time periods, immobility is not recommended mainly because of the restriction in ultimate motion which can be produced (Salter *et al.*, 1984) and because the mechanical properties of healing tissue such as ligaments and tendons under conditions of motion and slight tension, are superior to these properties under static conditions (Fronek *et al.*, 1983).

A reasonable design goal is that the knee brace provides an external guide to knee motion, compatable with the internal joint mechanics. However, while there are certain evident characteristics of motion such as the external rotation of the tibia as the knee is brought into extension, the motion can vary depending upon the activity and the external forces applied (Murphy *et al.*, 1985; Reuben *et al.*, 1986). Any motion path must fall within an envelope whose boundaries are defined by the laxity under realistic applied forces and moments (Hsieh and Walker, 1976; Markolf *et al.*, 1981; Fukubayashi *et al.*, 1982; Huiskes *et al.*, 1984; Shoemaker and Markolf, 1985). However, close to the extremes of laxity, certain of the ligaments and capsular soft tissues will be either tensed or very slack. Hence it may be useful for brace design to consider a central path of motion which could occur for example during normal gait, or during flexion–extension under quadriceps action, simulating rising from a chair or climbing a step (Chao *et al.*, 1983; Kurosawa *et al.*, 1985; Walker *et al.*, 1985; Reuben *et al.*, 1986). In those studies the most prominent motions apart from flexion–extension were found to be internal–external rotation and a–p translation. Neglect of these motions in a leg brace could have undesirable effects on internal knee mechanics, particularly on ligament length patterns. Studies of ligament lengths and tensions by many investigators (Huiskes *et al.*, 1984; Grood *et al.*, 1983; Girgis *et al.*, 1976; Trent *et al.*, 1976) have shown that different bands of the ligaments, particularly of the cruciates, have different length patterns during flexion–extension. The movements and shape of the menisci will also be affected by the motion. Normally, the medial meniscus appears to be relatively immobile due to the limited a–p excursion of the medial femoral condyle, while the lateral meniscus translates posteriorly with flexion. Ligamentous injuries resulting in abnormal motion can result in subsequent meniscal injuries. A further important consideration is that the general level of comfort of a leg brace and the brace's durability are likely to be a function of the mechanical compatability with the knee (Lewis *et al.*, 1984).

Our study was carried out to determine the effect of the design and placement of the external joints of a leg brace, on internal joint mechanics. The effects of motion incompatability on ligament length patterns, femoral–tibial contact points, meniscal movements and the likelihood of cuff-skin slippage were particularly considered.

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* Current address: Department of Biomedical Engineering, Institute of Orthopaedics, Stanmore, Middlesex, U.K.

MATERIALS AND METHODS

Geometry of ligaments and joint surfaces

Twenty-three intact fixed cadaveric knees were obtained. Radiography and direct observation ascertained the absence of any gross abnormalities. The musculature and joint capsule were removed, to clearly expose the ligaments. Metal pins 10 mm in length with spherical heads were placed at the femoral and tibial attachment points of the collateral and cruciate ligaments. The lateral collateral was marked with only one pin at each end. The broad medial collateral was marked with three pins on the femur and two on the tibia. For each cruciate attachment on the tibia, four pins were used to mark the four 'corners', antero-medial, antero-lateral, postero-medial and postero-lateral. The fibers from these pin locations were then traced to the femur where corresponding pins were placed.

In order to mount the knees to a reproducible axis system, the centers of the spherical surfaces of the posterior femoral condyles (Kurosawa *et al.*, 1985) were determined using radius gages. Using a jig, a Steinman pin was driven through this transverse axis. A cylindrical aluminium pot was cemented parallel to the shaft of the tibia and perpendicular to the transverse axis, and the knee was placed in extension in the jig. Key dimensions of the knees were measured (Mensch and Amstutz, 1976), in particular the medial–lateral width along the transverse axis which was used as the reference dimension for standardization. All coordinate data in the study was normalized to a medial–lateral dimension of 80 mm at the transverse axis.

Radiographs were taken with the center of the beam along the transverse axis (sagittal view) and perpendicular to the axis (frontal view) (Fig. 1). The radiographs were then placed on a Talos digitizer and the ligament attachment points were digitized into a Chromatics CGC 7900 colorgraphics computer. The actual xyz coordinates of the attachment points were computed, using the known distances between the X-ray source, knee origin and film plate. Using the same axis system, and with the knees maintained in extension, each knee was placed in a plastic box and embedded in Pedilen polyurethane foam. Twenty-four equally spaced sagittal slices were cut using a band saw. The 25 sections were copied at 1:1 and the profiles of the femur and tibia were digitized. Up to 40 points were used for each section with a greater density of points on the cartilage outline. These latter points were typically spaced apart by 3–4 mm.

Motion analysis

The objective of this part of the study was to impose various motions on the knees, and determine the predicted overlap or separation of the femoral–tibial condyles, the ligament length patterns and the femoral–tibial contact point locations. In previous studies (Kurosawa *et al.*, 1985; Walker *et al.*, 1985) 'average knee motion' was reported from studies of flexion–extension under quadriceps action. The axis system was defined the same as above. The femur and tibia were assumed to have coincident axes in extension. The relative motion between the femur and the tibia was expressed as three successive Eulerian rotations in the order flexion, varus, internal rotation and three translations of the femoral origin relative to the axis system fixed in the tibia. A more recent study has been carried out using an improved technique where the knees were flexed and extended continuously using a motor to lengthen and shorten the quadriceps. (Reuben *et al.*, 1986). The results for average knee motion confirmed the data of Kurosawa with only minor differences. Five degrees of freedom were fitted to equations as a function of flexion angle:

$$\text{VARUS} = (0.0791 \times F) - (5.733\,E-04 \times F^2) - (7.682\,E-06 \times F^3) + (5.759\,E-08 \times F^4) \quad (1)$$

$$\text{INTROT} = (0.3695 \times F) - (2.958\,E-03 \times F^2) + (7.666\,E-06 \times F^3) \quad (2)$$

$$x\text{DIS} = 0.0 \quad (3)$$

$$y\text{DIS} = -(0.0683 \times F) + (8.804\,E-04 \times F^2) - (3.750\,E-06 \times F^3) \quad (4)$$

$$z\text{DIS} = -(0.1283 \times F) + (4.796\,E-04 \times F^2) \quad (5)$$

F = angle of flexion, extension being zero. Units are degrees and mm.

The femoral geometry, the tibial geometry and the ligament attachment coordinates were stored as data arrays in the computer. For the calculations, all knees were standardised to an 80 mm medial–lateral width by multiplying all coordinates by $80/ML$ where ML was the actual ML width of that knee. Average knee motion, or another defined motion, was imposed on the knee. If the coordinate of any femoral point with respect to the axis system at zero degrees flexion was x_0, y_0, z_0 then the transformed coordinates after flexion F were obtained by multiplying x_0, y_0, z_0 by a 4×4 matrix, which included three angles and three displacements $(C_1) = (M)(C_0)$ (equation 6). This matrix has been described elsewhere (Beggs, 1967; Kurosawa *et al.*, 1985). Abnormal motions were simulated by modification of equations (1)–(5). For example, to simulate the elimination of internal–external rotation, equation (2) above becomes: INTROT = 0.0. The abnormal motions considered were 'rotation eliminated' (equivalent to a 'polycentric hinge') and 'rotation and a–p eliminated' (almost equivalent to a fixed axis hinge). A different type of abnormal motion was modelled by assuming average knee motion, but with the origin at zero degrees flexion offset by specified amounts along the y and z axes. This simulated an external hinge with average knee motion, but displaced from its 'correct' location on the knee. The following were calculated as a function of the type of motion, for a flexion range of 0–120°:

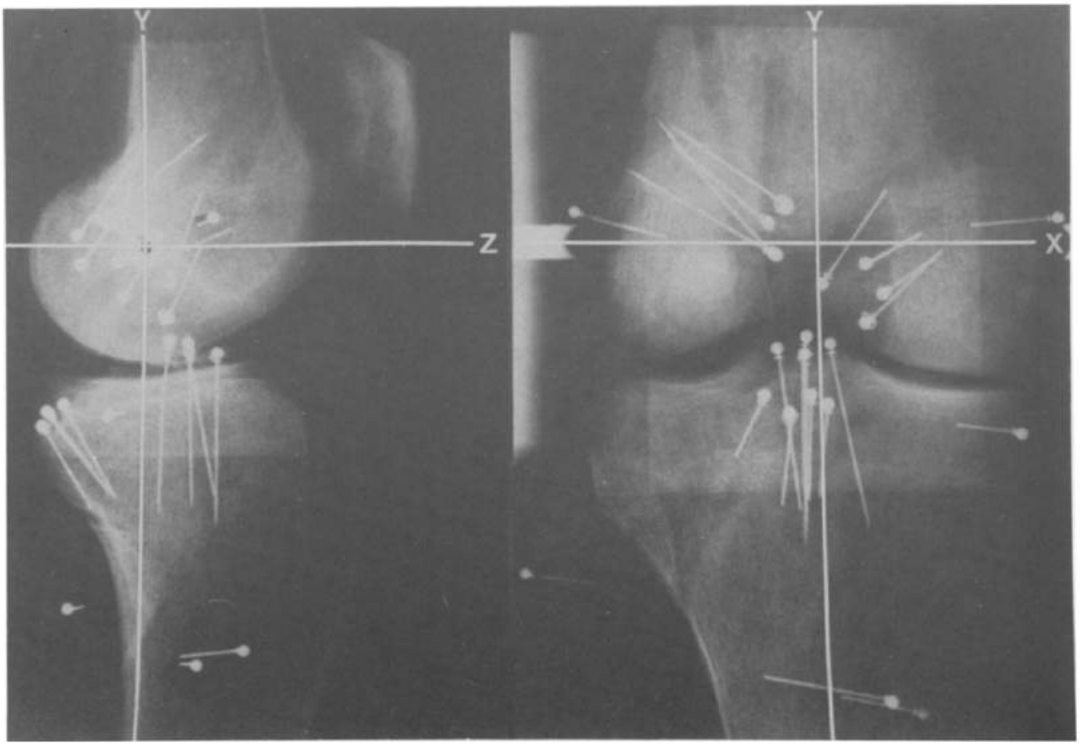

Fig. 1. Sagittal and frontal radiographs of a typical knee specimen. The pins mark the ligament attachment points. The axis system is shown. The X-ray beam was centered at the origin for each view.

(a) The gaps or overlaps (called contact errors) between the femoral and tibial condylar surfaces on the lateral and medial sides and the correction in the frontal plane position of the femur on the tibia to produce point contacts.

(b) The coordinates of the lateral and medial femoral–tibial contact points in the transverse (xz) plane.

(c) The lengths between the femoral and tibial ligament attachment points, with the lateral and medial condyles in point contact.

Computer algorithms

A computer program was written which performed the following sequence of algorithms

(i) To determine the contact errors between the femoral and tibial surfaces, consider a point F on a femoral section (Fig. 2). The tibial sections medial and lateral to this point, [sections P and $(P+1)$] were determined. Similarly, the points posterior and anterior to F on the tibial sections P and $(P+1)$ were determined. These points were J and K on section P and L and M on section $(P+1)$. Points G and H were defined as lying on line JK and LM and with the same z value as point F. Finally, E was the intersection of the vertical (parallel to the y-axis) line through F with line GH. The distance FE was then defined as the contact error between the femoral point and the tibial surface. A point search routine was written to determine the minimum contact errors on the lateral and medial sides. A positive contact error represented a gap or separation between the femoral and tibial surfaces: a negative contact error, an overlap.

The resolution was limited in two ways. Because of linear extrapolation on the tibial surface, E will in general not be located on the surface: however, due to the small curvatures involved, the errors were calculated to be less than 0.2 mm. Because only the digitized points on the femoral surface were analyzed, the resolution in the xz plane was only ± 2 mm in any direction.

(ii) Assume that the minimum contact errors calculated initially for the lateral and medial sides are LATMIN and MEDMIN respectively (Fig. 3). The xy axes are fixed in the tibia. The femur can be brought into contact by a rotation about the origin and by translations along the x and y axes. The aim was to minimize the medial–lateral (along x) translations at the contact points. (The calculations are given in the Appendix.) The rotation and translations were then added to the original values and the femur was retransformed according to equations (1–6). However, in a few cases, this correction resulted in different femoral points becoming closest to the tibial surface and non-zero contact errors. In this case, the program iterated until the contact errors were less than a specified minimum, set at 0.5 mm.

(iii) In all of the knees, there were small contact errors at 0° of flexion on the lateral and medial sides (mean values, lateral 1.70 mm., medial 0.11 mm) due to experimental error, but also probably due to actual separation between the condyles in some knees when they were being embedded. These initial contact errors were eliminated by calculating the above frontal plane corrections and transforming the tibial data array. This in effect, adjusted the tibia so that it contacted the femur on the lateral and medial sides.

(iv) At angles of flexion beyond 0°, the femoral points closest to the tibial surface, were taken to represent the femoral–tibial contact points. The coordinates in the transverse xz plane were thus determined.

(v) After carrying out the iterations described in (ii) above to bring the femoral and tibial condyles into contact, the final values of VARUS, INTROT, xDIS,

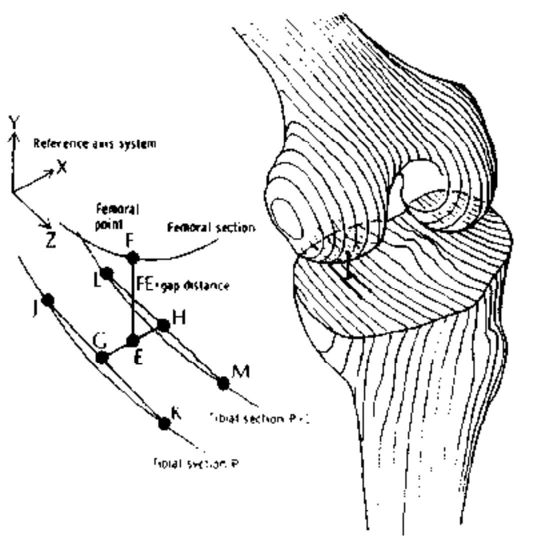

Fig. 2. Determination of contact error between the femoral and tibial surfaces. F is a point on a femoral section. The line FE is parallel to the Y-axis. Distance FE is defined as the contact error (in this case a gap distance).

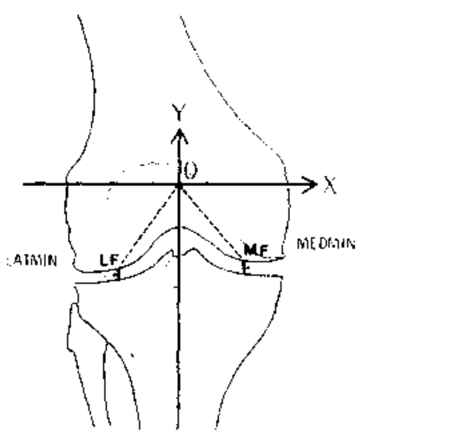

Fig. 3. The contact errors are LATMIN and MEDMIN. To close the gaps, the femur was rotated about 0 and translated along x and y. Details are given in the appendix.

yDIS and zDIS were known. The femoral ligament attachment coordinates were transformed using equations (1)–(6). The ligament fiber lengths were then calculated from the equation

$$L = \mathrm{SQRT}[(x_F - x_T)^2 + (y_F - y_T)^2 + (z_F - z_T)^2]$$

where suffices F and T refer to the femur and tibia respectively.

RESULTS

(a) *Contact errors*

The magnitudes of the means and S.D. of the contact errors for average knee motion were calculated (Fig. 4). At 15° of flexion and higher, the mean contact errors on the lateral side ranged from -1.81 to -0.44 mm indicating an overlap of the femoral and tibial surfaces. On the medial side, the errors were comparable in magnitude at 0.58–1.91 mm indicating a separation. The S.D. ranged from 1.71 to 3.16 mm. The contact errors were considered to be due to the variations in geometry from knee to knee and to the difference between the actual motion a particular knee would show from average knee motion. In the Reuben *et al.* study (1986) varus–valgus was found to be particularly variable. A 2° varus or valgus rotation would produce a gap of about 2 mm on one side of the knee.

Eliminating both internal–external rotation and anterior–posterior displacement to simulate a fixed hinge produced differences in contact errors compared with average motion. On the medial side however, the contact errors were very similar to those of average motion, which may be because the medial tibial plateau lies close to a horizontal plane and changes of rotation or a–p displacement of the femur would not grossly affect the contact situation. On the lateral side, the mean contact error increased steadily with flexion to more than 4 mm. This is likely to be due to the posterior slope of the lateral tibial plateau, combined with lack of rollback of the contact points.

When the axis of average knee motion was offset by 5 mm, the mean contact errors were as high as 8 mm (Fig. 5). For an upwards or downwards offset, the contact errors were similar to those for average motion up to 60° flexion, but thereafter became much larger. For an anterior or posterior offset, the contact errors were larger than for average motion from 30° of flexion. When the axis offset was upwards or posterior, the contact errors increased positively with flexion, indicating femoral–tibial gaps. The reverse was the case for downwards or anterior axis offset.

The S.D. for the offset data fell mostly in the range of 1–4 mm but in a few instances it was as high as 7 mm. Despite this, all of the contact errors (except four) for offset axes were significantly different from the zero offset data, when using a paired t-test. This indicated consistent relative data from knee to knee.

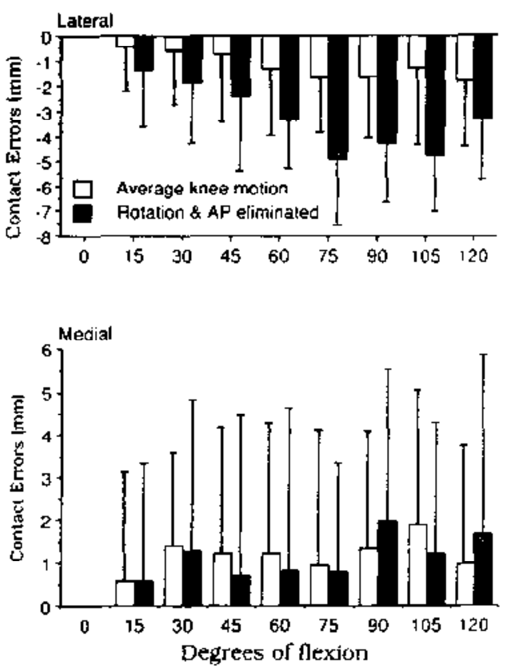

Fig. 4. The contact errors on the lateral and medial sides, for average knee motion, and for 'rotation and a–p eliminated' (approximating a fixed axis). Differences were significant at $p<0.01$ on the lateral side in a paired t-test.

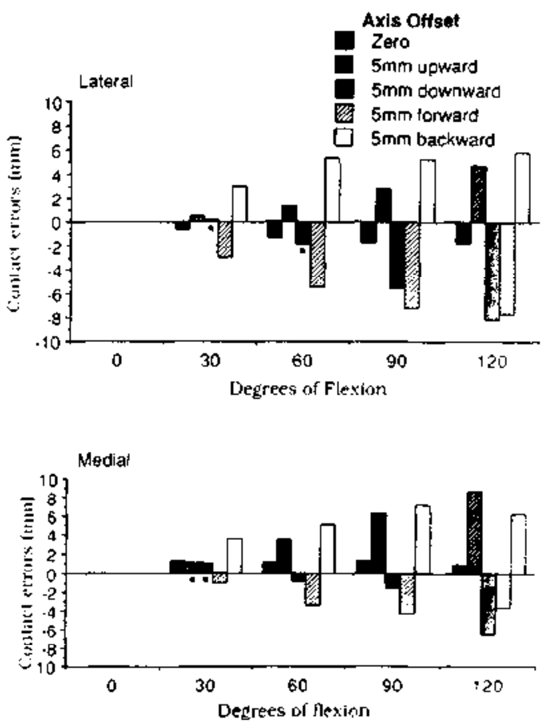

Fig. 5. The mean contact errors for errors in placement of a hinge producing average knee motion. Zero offset is correct placement. Differences in offset data were different from zero offset data at $p<0.01$ except when marked by an asterisk. A paired t-test was used for analysis.

(b) *Contact point locations*

For average knee motion, the contact points on the lateral and medial sides moved generally posteriorly with flexion (Fig. 6). The rate of the posterior movement decreased with angle of flexion. On the lateral side, the initial contact point was close to the center in the a–p direction at zero flexion, moved backwards by 10 mm (for the average sized knee) by 30° flexion, moved a further 4 mm to 45° flexion and thereafter remained constant. On the medial side, the contact point at 0° flexion was 30% from the anterior edge, and moved uniformly backwards by 13 mm up to 45° flexion, after which it remained constant. The standard deviations in the contact point locations were approximately ± 3 mm in the a–p direction and ± 5 mm in the m–l direction.

When internal–external rotation was eliminated, the lateral excursion decreased by 30% and the medial excursion increased by 30%. When anterior–posterior displacement was eliminated, the posterior rollback on the lateral side was again reduced by 30% but there was an even larger reduction on the medial side of 40%. When both rotation and a–p displacement were eliminated, the rollback on the lateral side was further reduced to less than 4 mm but on the medial side the contact point locations reverted to almost normal.

(c) *Ligament length patterns*

The means and S.D. of the ligament-length patterns for average knee motion are shown in Fig. 7. The fiber lengths are plotted as ratios, relative to the lengths at 0° flexion. The results for average motion are strikingly similar to those published by Huiskes *et al.* (1984). The anterior fibers of the anterior cruciate were of constant length throughout flexion, while the posterior fibers decreased in length by 15%. The opposite was the case for the posterior cruciate: the posterior fibers remained within 10% of their initial length, but beyond 30° flexion the anterior fibers steadily increased in length to reach a 25% increase by 120° flexion. The anterior fibers of the medial collateral remained at the same length throughout flexion, whereas the posterior fibers decreased by 10% during flexion. The lateral collateral decreased in length by 20%.

When internal–external rotation was eliminated, there were no significant changes in the ligament length patterns. However, when anterior–posterior displacement was eliminated, the strain in the anterior cruciate decreased by almost 20%, while there was a similar increase in the posterior cruciate. This produced large strain differences of up to 40% in certain of the fibers. The changes in the collateral ligaments were minimal. There were no additional changes in the length patterns when both rotation and a–p were eliminated.

For an upwards or posterior offset of 5 mm in the axis of the average knee motion, the changes in the length patterns were similar: the strains in the posterior cruciate were decreased by about 15%, resulting in less overall strains during the flexion range. There was a further loosening of about 15% in the anterior cruciate. There was also increased loosening of the collaterals with flexion. For a downwards or anterior

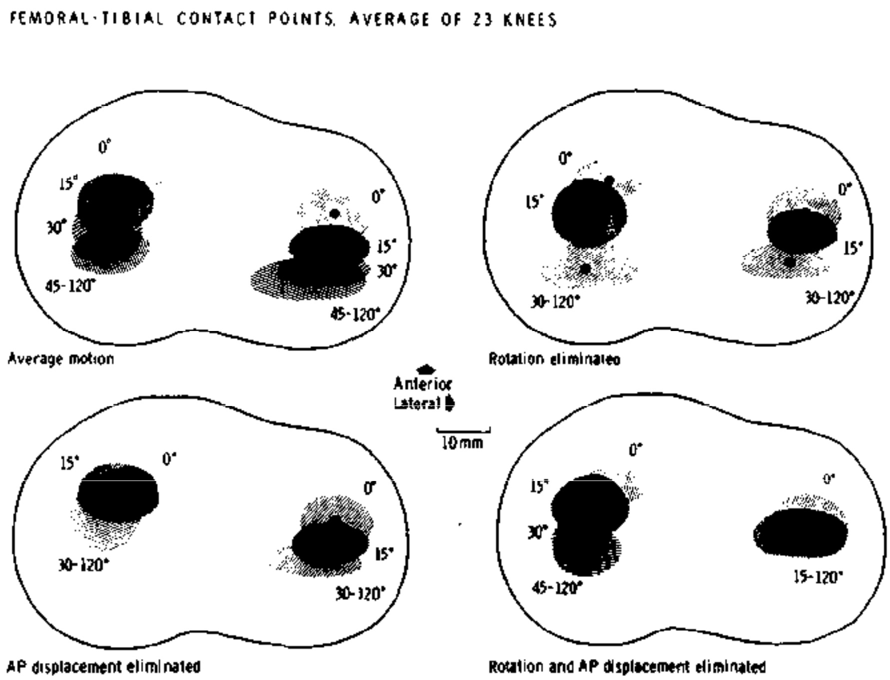

Fig. 6. The femoral–tibial contact point locations for average knee motion and for three abnormal motions. The mean contact points are black dots: the S.D. are shaded.

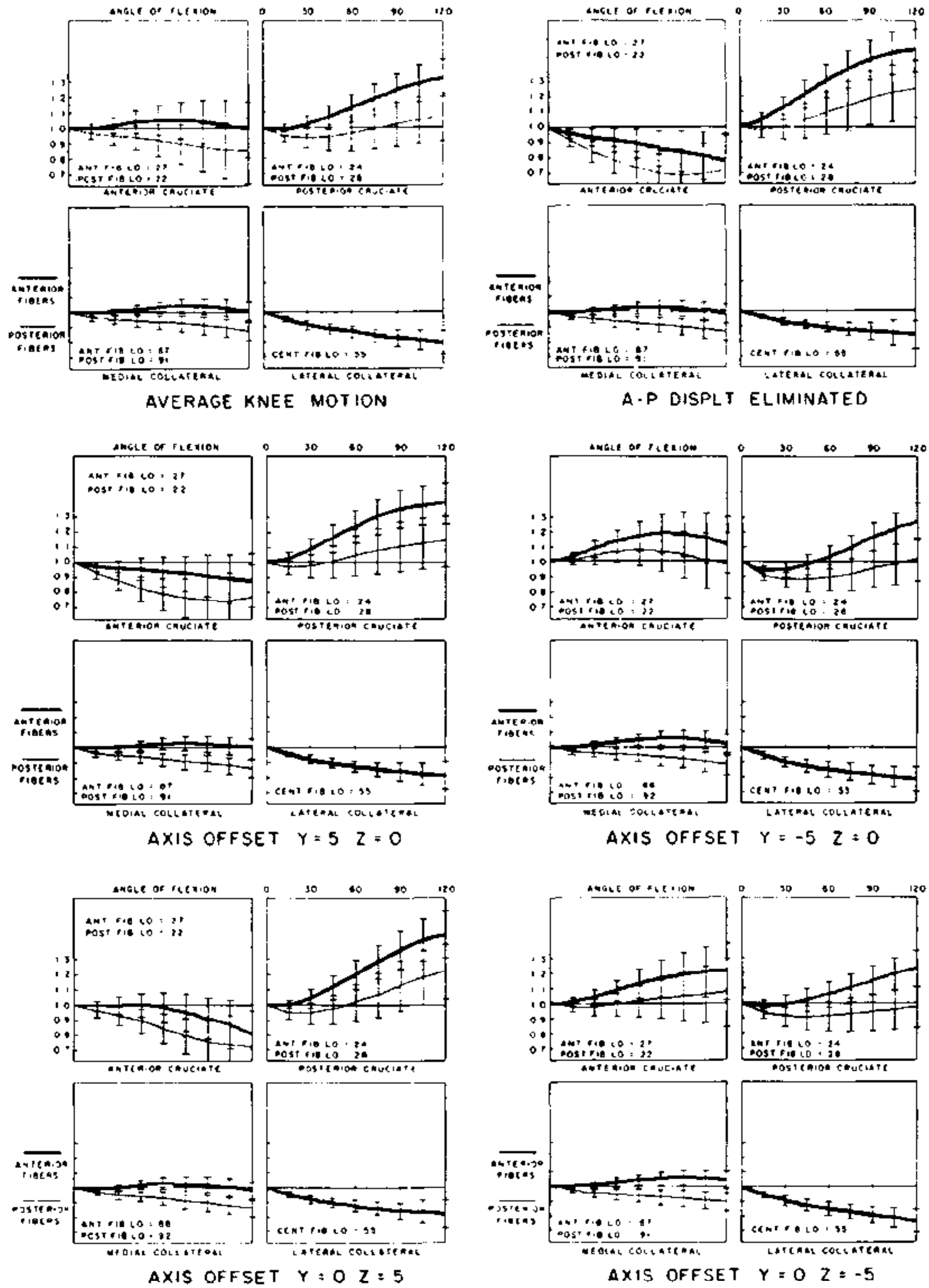

Fig. 7. The ligament length patterns for average knee motion, for a–p displacement eliminated, and for offset of average knee motion. The mean and S.D. data are shown for the 23 knees.

offset of 5 mm, the effect of the PCL was reversed, namely an increase in strain, resulting in a large total length increase of about 50% in the anterior fibers. On the other hand there were only small changes in the ACL and in the collaterals.

DISCUSSION

The efficacy of knee braces in controlling knee motion has been subject to question recently by the realization that at this time, many of the models have little or no supporting data and the data that is available suggests that the braces exert only partial control and then only under force conditions that are less than in many physiological activities (Knutzen *et al.*, 1984; AAOS, 1985). There are many possible reasons, but the most important are likely to be the difficulty of obtaining a rigid connection between the brace and the bones, the compliance and changing shape of the soft tissue, the general flexibility of the braces themselves and the slippage which occurs between the cuffs and the skin. There is little quantitative data however on these various factors.

Superimposed on the above factors are the kinematic characteristics of the external hinges and their placement relative to the bone geometry. These aspects were the subject of our study. The underlying theme is, if a brace could be rigidly connected to the bones, what would be the effect of different hinge motions and placements on the joint mechanics? Additionally, would an off-the-shelf hinge incorporating average knee motion produce satisfactory mechanics in a population of knees which would have a variety of geometry and normal motion paths? The rationale for the method relies on the acceptance of the concept of an 'average knee motion', or even on an average motion for a particular knee. The motion was measured under the action of the quadriceps with the knee under compressive load, on the assumption that the knee would undergo a motion path reasonably in the center of the laxity range. The average motion from the group of knees was obtained by averaging the curves for each degree of freedom in turn as a function of flexion angle.

Not surprisingly, the contact errors for the 23 knees investigated were small when average knee motion was imposed on them. The tibial plateaus are oriented close to the horizontal plane and the contact errors could be accounted for by small variations in tibial plateau tilt and femoral radii of curvature. Such errors would be readily absorbed in shear in the soft tissue around the joint. However, elimination of rotation and a–p displacement in the hinges (fixed axis motion) led to large errors on the lateral side at the higher flexion angles. Average motion which was offset only 5 mm from the correct axis, led to even larger contact errors which were highest for forward and backward offsets, and least for upward and downward offsets, up to 90° of flexion. These errors would of course not occur in practice; a gap would be eliminated under joint load, while a condylar overlap would be prevented by the joint surfaces.

Abnormal motions resulted in differences in the positions of the femoral–tibial contact points. This would be unlikely to lead to damage of the joint surfaces, but there may be a risk of damage to the menisci by straining of their attachments to the capsule. Meniscal positions were found to be abnormal for all motions which included elimination with a normal motion and location pattern of a–p displacement or internal–external rotation only in the rotation eliminated case.

Ligament length patterns were considerably altered by abnormal motion or axis location. Tension in the posterior cruciate would be particularly increased for a–p eliminated or distal or posterior axis placement.

The effects of these theoretical contact errors, contact locations, and ligament length changes, will be a compromise between the motion which the knee would naturally undergo and the motion which the external joint attempts to impose. The result will be one or more of shear in the tissues between the cuffs and the bones, slippage between the cuffs and the skin, superimposed forces and moments across the joint, and equal and opposite forces and moments across the external hinges.

Such forces acting on the hinges were determined experimentally and found to correlate with inaccurate motion and placement (Regalbuto *et al.*, 1987). Pistoning forces were also measured as a function of placement by Lew *et al.* (1984); and Lewis *et al.* (1984). Differences in the pistoning forces were masked by variations of location due to repeated donning and doffing, which is consistent with our result of large changes in the mechanics due to only 5 mm of hinge misplacement.

It is concluded that if external joints are attached to a normal knee, for prophylactic or functional application, accurate placement of external joints with built-in average knee motion should reduce unwanted forces across the natural knee and the external joints, and reduce cuff–skin slippage and soft tissue shear. If the external joints are applied to knees with abnormal laxity due to ligament injuries, or to postoperative knees, built-in average knee motion, or another motion for required ligament length patterns, together with accurate placement, will satisfy the goals if the natural joint follows the motion of the external joint. This might be feasible under passive motion conditions. However, under functional conditions, the compliance of the soft tissue between the external joints and the bones is likely to lead to a reduction in efficacy. The maximum degree to which an external joint can accurately control knee motion under these functional conditions needs further research.

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REFERENCES

American Academy of Orthopaedic Surgeons (1985). *Knee Braces* Seminar Report August 17–18, 1984. Chicago, IL, U.S.A.

Beggs, J. S. (1967) *Advanced Mechanism*. Macmillan Company, New York.

Butler, P. B., Evans, G. A., Rose, G. K. and Patrick, J. H. (1983) A review of selected knee orthoses. *Br. J. Rheumatol.* 22, 109

Chao, E. Y. S., Laughman, R. K. Scheider, E. and Stauffer, R. (1983) Normative data of the knee joint motion and ground reaction forces in adult level walking. *J. Biomechanics* **16**, 219.

Fronek, J., Frank, C., Amiel, D., Woo, S. L.-Y., Coutts, R. D. and Akeson, W. H. (1983) The effect of intermittent passive motion on the healing of the medial collateral ligament. *Proceedings of the Orthopedic Research Society*, Anaheim, CA, p. 31.

Fukubayashi, T., Torzilli, P. A., Sherman, M. F. and Warren, R. F. (1982) An *in vitro* biochemical evaluation of anterior-posterior motion of the knee. *J. Bone Jt Surg.* **61-A**, 258.

Girgis, F. G., Marshall, J. L. and Al Monajem, A. R. S. (1975) The crutiate ligaments of the knee joint. *Clin. Orthop.* **106**, 216.

Grood, E. S., Hefzy, M. S., Butler, D. L., Suntay, W. J., Siegel, M. G. and Noyes, F. R. (1983) On the placement and the initial tension of anterior cruciate ligament substitutes. *Proceedings of the Orthopedic Research Society*, Anaheim, CA, p. 92.

Hsieh, H. H. and Walker, P. S. (1976) Stabilizing mechanisms of the loaded and unloaded knee joint. *J. Bone Jt Surg.* **58-A**, 87.

Huiskes, R., Blankevoort, L., van Dijk, R., de Lange, A. and van Rens Th., J. G. (1984) Ligament deformation patterns in passive knee joint motions. *Advances in Bioengineering*, (Edited by R. Spilker). American Society of Mechanical Engineers, New York.

Huiskes, R., de Lange, A. and Hamer, A. (1984) The influence of loading and meniscectomy on the 3-D freedom of motion characteristics of the knee joint. *Proceedings of the Orthopedic Research Society*, Atlanta, GA, p. 27.

Knutzen, K. M., Bates, B. T. and Hamill, J. (1984) Knee brace influences on the tibial rotation and torque patterns of the surgical limb. *J. Orthop. Sports Phys. Ther.* **6**, 116.

Kurosawa, H., Walker, P. S., Abe, S., Garg, A. and Hunter, T. (1985) Geometry and motion of the knee for implant and orthotic design. *J. Biomechanics* **18**, 847.

Lew, W. D., Patrnchak, C. M., Lewis, J. L. and Schmidt, J. (1984) A comparison of pistoning forces in orthotic knee joints. *Orthot. Prosthet.* **36**, 85.

Lewis, J. L., Lew, W. D., Patrnchak, C. M. and Shybut, G. T. (1984) A new concept in orthotics joint design. *Orthot. Prosthet.* **37**, 15.

Lewis, J. L., Lew, W. D., Stulberg, S. D., Patrnchak, C. M. and Shybut, G. T. (1984) A new concept in orthotics—Part II. the complete orthosis. *Orthot. Prosthet.* **38**, 13.

Markolf, K. L., Bargar, W. L., Schoemaker, S. C. and Amstutz, H. C. (1981) The role of the joint load in knee stability. *J. Bone Jt Surg.* **60-A**, 570.

Mensch, J. S. and Amstutz, H. C. (1975) Knee morphology as a guide to knee replacement. *Clin. Orthop.* **112**, 231.

Murphy, M. C., Zarins, B., Jasty, M. and Mann, R. W. (1985) *in vivo* measurement of the 3-dimensional skeletal motion at the normal knee. *Proceedings of the Orthopedic Research Society*, Las Vegas, NA, p. 142.

Noyes, F. R., Grood, E. S., Butler, D. L. and Malek, M. (1980) Clinical laxity tests and functional stability of the knee: biomechanical concepts *Clin. Orthop.* **146**, 84.

Regalbuto M. A., Schrager R., Rovick, J. S. and Walker, P. S. (1987) The effectiveness of knee braces as a function of hinge design and placement. *Proceedings of the Orthopedic Research Society*, San Francisco, CA, p. 246.

Reuben, J. D., Rovick, J. S., Walker, P. S. and Schrager, R. J. (1986) Three-dimensional kinematics of normal and cruciate deficient knees–a dynamic *in vitro* experiment. *Proceedings of the Orthopedic Research Society*, New Orleans, LA.

Salter, R. B., Hamilton, H. W., Wedge, J. H., Tile, M., Torode, I. P., O'Driscoll, S. W., Murnaghan, J. J. and Saringer, J. H. (1984) Clinical application of basic research on continuous passive motion for disorders and injuries of synovial joints: a preliminary report of a feasibility study. *J. Orthop. Res.* **1**, 325.

Shoemaker, S. C. and Markolf, K. L. (1985) Effects of joint load on the stiffness and laxity of ligament-deficient knees. *J. Bone Jt Surg.* **67-A**, 136.

Trent, P. S., Walker, P. S., and Wolf, B. (1976) Ligament length patterns, strength, and rotational axes of the knee joint. *Clin. Orthop.* **117**, 263.

Walker, P. S., Kurosawa, H., Rovick, J. S. and Zimmerman, R. A. (1985) External knee joint design based on normal motion *J. Rehab.* **22**, 9.

APPENDIX. CORRECTION OF THE CONTACT ERRORS (FIG. 3)

LF and MF are the femoral points on lateral and medial sides respectively.

The corrections to the femur to achieve zero contact error, minimizing the medial–lateral (along x) displacements of the contact points are: ϕ, clockwise about O; yCOR along y axis; xCOR along x axis.

$x\mathrm{LF}$ = x-coordinate of point LF etc.

$x'\mathrm{LF}$ = x-coordinate of point LF after rotation ϕ about 0 etc.

Since

$yLF = yMF$ and $|yLF - yMF| \ll |xLF - xMF|$:

$$\phi = \frac{\mathrm{MEDMIN} - \mathrm{LATMIN}}{xMF - xLF}$$

$$x'LF = xLF.\cos\phi + yLF.\sin\phi$$

$$y'LF = -xLF.\sin\phi + yLF.\cos\phi$$

$$x'MF = xMF.\cos\phi + yMF.\sin\phi$$

$$y'MF = -xMF.\sin\phi + yMF.\cos\phi$$

$$x\mathrm{COR} = 1/2[(DMF - x'MF) + (xLF - x'LF)]$$

$$y\mathrm{COR} = 1/2[[\mathrm{LATMIN} + (y'LF \quad yLF)] + [\mathrm{MEDMIN} + (y'MF - yMF)]]$$