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A novel total knee replacement by rolling articulating surfaces. In vivo functional measurements and tests

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The purposes of the paper were as follows: to show the fundamental functional differences between the natural knee and common total knee replacements (TKR), to describe the ideas on how main properties of the natural knee can be adopted by a novel TKR and to present some main biomechanical functions of this TKR. By analyzing the morphology of the articulating surfaces and the kinematics of the natural knee the design of the novel TKR was developed. The use was made of the test procedures established in vitro and of lateral X-ray photographs as well as fluoroscopy in vivo. The function of the novel TKR is comparable to that of the natural knee joint in terms of kinematics (roll/slide behaviour), loads of the articulating surfaces (diminished shear loads), stability and leeway under external impacts, reduction of the load in the patellofemoral joint, and ligament balancing.

Key words: *total knee replacement, rolling articulating surfaces, loads of articulating surfaces, in vivo tests and measurements, AEQUOS-G1 prosthesis*

1. Common total knee replacement versus natural tibiofemoral joint

1.1. Total knee replacement (TKR)

Until recently, in almost all TKR available on the market, the sagittal profiles of the medial and lateral articulating surfaces have been identical in terms of design and function. In particular, the rotational axis defined by the articulating surface in the lateral femoral compartment and the rotational axis defined by the articulating surface in the medial femoral compartment lie always one upon the other. Implanted, this geometric feature causes the following kinematic TKR characteristics:

1. During knee flexion the lateral and the medial joint contact remains practically stationary on the tibial articulating surfaces and consequently the tibia plateau slides along both femoral articulating surfaces in the same way. The friction mechanism in knee flexion is specified by sliding friction and by static friction in the case of the reversal of movement. Thus, in particular during the stance phase of gait (figure 1, [2]), considerable sliding friction must occur because of the high joint load and also high static friction, since the knee bending movement alters its sign three times in this gait phase. At these points of reversing, static friction occurs normally which produces detrimental shear stresses at the interfaces between implants and bone. Up to now, these friction problems are exclusively obviated by suitable material (lubricate) pairings.

2. In the course of the movement described in point 1, the angle β between the patella tendon and the "tibial plateau" remains practically stationary as shown by PANDIT et al. [1]. In flexion, the angle between the force line of the quadriceps femoris and the force line of the patella tendon becomes sharper

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causing a steep increase of the contact load in the patellofemoral joint.

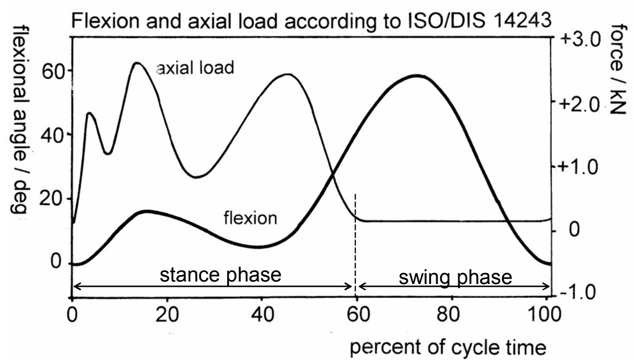

Fig. 1. Gait cycle: during the stance phase the compressive load in the knee joint may be more than three times as great as the body weight. The flexion angle approaches 30°

1.2. Natural tibiofemoral joint (TFJ)

Morphology: In the natural knee, the curvature morphology of the lateral tibial articulating surface is obviously different from that of the medial one as sagittal sections show (figure 2): a) the sagittal tibial contours are convexly curved in the lateral compartment, but concavely curved in the medial one; b) the axes, defined by the femoral sagittal contours on the lateral or on the medial side, do *not* lie one upon the other; the medial axis lies somewhat shifted to the anterior[1].

Kinematics: As early as in 1836, the WEBER brothers [3] reported that during human knee flexion the tibial articulating surfaces both slid and rolled along the femoral condyle. In his textbook in 1907, FISCHER [4] summarized the in vitro measurements of LANGER (1858) [5], the in vivo measurements of BRAUNE & FISCHER (1891) [6], and the in vivo radiographic measurements of ZUPPINGER (1904) [7] on knee movement and precised the statements of the Weber brothers: from full extension the articulating surfaces should initially roll and then predominantly slide medially for angles of flexion >15° and laterally >20°, causing both articulating contacts to become fixed on the tibia plateau. Fischer also explained that the IRA (instantaneous axis of rotation), initially in a distal position, would migrate with increasing flexion angle into the centre of the femoral condyle. WALKER et al. (1988) [8] confirmed the initial posterior migration of both contacts on the tibial plateau and established that they would become fixed after approx. 25° flexion. The computer model by WISMANS et al. (1980) [9], which used empirically recorded articulating surface morphology, produced similar results. In 2005, LI et al. [10] presented computer models, which were based on in vivo measurements by Dual-Orthogonal Fluoroscopy and Magnetic Resonance. They reported that up to 30° the contact would migrate to the posterior in the medial as well as in the lateral compartment. In 2001 and 2004, PINSKEROVA et al. [11], [12] presented in vivo MRT sections through TFJ in several flexion positions. In evaluating the picture series, the authors summarized that "there is no rolling at the contact area". This conclusion was incorrect because the evaluating methods were mathematically and physically oversimplified. Applying correct mathematical procedures in order to evaluate exactly the same data we got contrary results and could show that in the initial range of flexion a considerable portion of rolling occurred (tables 1, 2). Considering this unique roll-/sliding behaviour in the aspect of the human gait one can assess the perfect

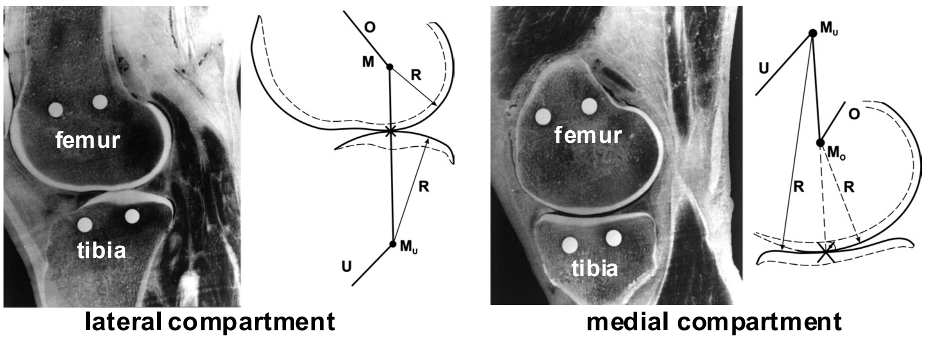

Fig. 2. Sagittal sections of the lateral and the medial compartments of a human knee joint (autopsy material). At the right side the contours of the articulating surfaces are traced. In each compartment, two rotational axes (*M*) are morphologically defined. Note: the lateral tibial articulating surface is convexly curved

[1] Unpublished our own measurements.

Table 1. Results of our re-analyses of the data reported by PINSKEROVA et al. [11].
The mean rolling/sliding relation approached unity at low angles of flexion: predominant rolling!

Rolling/sliding relation	Range of flexion	Medial compartment	Lateral compartment
$\bar{\rho}$	5°–10°	0.68	0.81
$\bar{\rho}$	45°–90°	0.12	0.22

Table 2. Results of our re-analyses of the data reported by PINSKEROVA et al. [12].
In vivo, the natural knee initially showed almost perfect rolling, especially under load.
$\bar{\rho}$ = 1.24 means that the portion of gliding (19%) is due to a sliding of the femur along the tibial surface

Rolling/sliding relation	Range of flexion	Medial compartment			Lateral compartment		
		Loaded	Unloaded	Cadaveric	Loaded	Unloaded	Cadaveric
$\bar{\rho}$	0°–20°	0.96	0.80	0.63	1.24	0.17	0.37
$\bar{\rho}$	45°–90°	0.10	0.09	0.08	0.20	0.43	0.43

mechanics of the human knee. During the stance phase, when the ground reaction force heavily loads the TFG, its articulating surfaces roll! Nature solved the problem of friction not only by suitable lubricate pairings, but also kinematically by rolling instead of sliding. As already mentioned above, the angular velocity of flexion varies three times its sign at 0°, 5°, and 18°. But static friction cannot play any role since the natural TFG is rolling at these flexion angles.

Patella loading: According to PANDIT et al. [1] the patella tendon angle β posteriorly swivels by 0.3° per 1° flexion under load. This fact limits the flexural load and especially the contact load in the femoropatellar joint, because the angle between the force line of the quadriceps femoris and the line of the patella tendon is increased by the alteration of the angle β.

2. General design of a TKR with initially rolling articulating surfaces

To implement rolling at small flexion angles, the lateral and medial compartments were asymmetrically designed. In sagittal direction, the lateral tibial articulating surface was convexly shaped, as opposed to the concavely formed medial surface. The medial femoral articulating surface has been shifted some millimetres to the anterior and superior compared with the almost identically shaped lateral femoral surface (figure 3). Figure 4 show the AEQUOS-G1 prosthesis.

In the presence of a compressive joint force, the guidance by this system of articulating surfaces leads to a constrained motion in flexion/extension. It is equivalent to that of a four-bar-chain. The four-hinge axes of the chain are defined by the rational axes of the four articulating surfaces (figure 3). According to the measures of the bars the four-bar-chain represents a double throttle crank. For small flexional angles its instantaneous rotational axis (IRA) could be positioned closely to the contacts of the articulating surfaces by optimizing the parameters of the gear so that predominant rolling could be expected (figure 5). In the course of the further increasing flexional angle, the IRA migrates to the centre of the femoral condyle yielding a predominant sliding of the articulating surfaces during the swing phase of the gait.

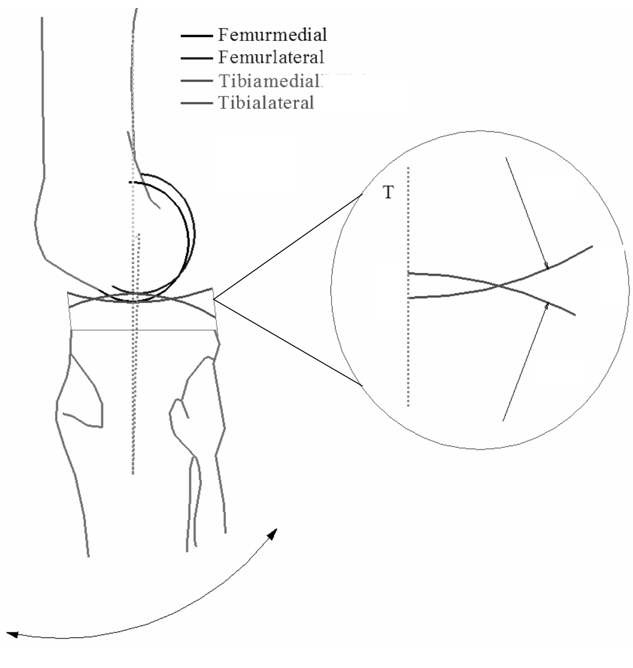

Fig. 3. Sagittal view of the incongruous contours of the articulating surfaces in the AEQUOS-G1

Stability and leeway: Frontal sections of the prosthesis demonstrate that also the transversal tibial and femoral curvature radii do not fit and that the "point" contacts between the tibial and femoral surfaces are positioned at the tibial inclines (figure 6).

Fig. 4. AEQOUS-G1: the tibia inlay must be cooled down to about 4 °C and then inserted in a metal frame. After the warming up to the body temperature the inlay is fixed

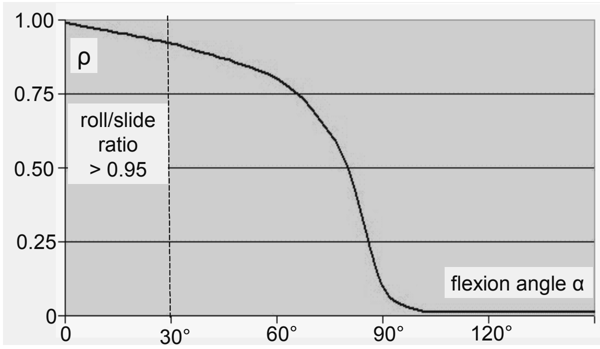

Fig. 5. Calculated roll/slide ratio of the AEQUOS-G1 at the medial side

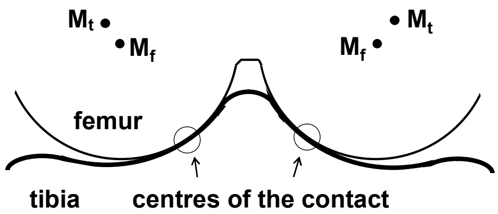

Fig. 6. In the AEQUOS-G1, the contacts are positioned at the tibial inclines. M_t, M_f – the centres of the curvature of the tibial and femoral contours

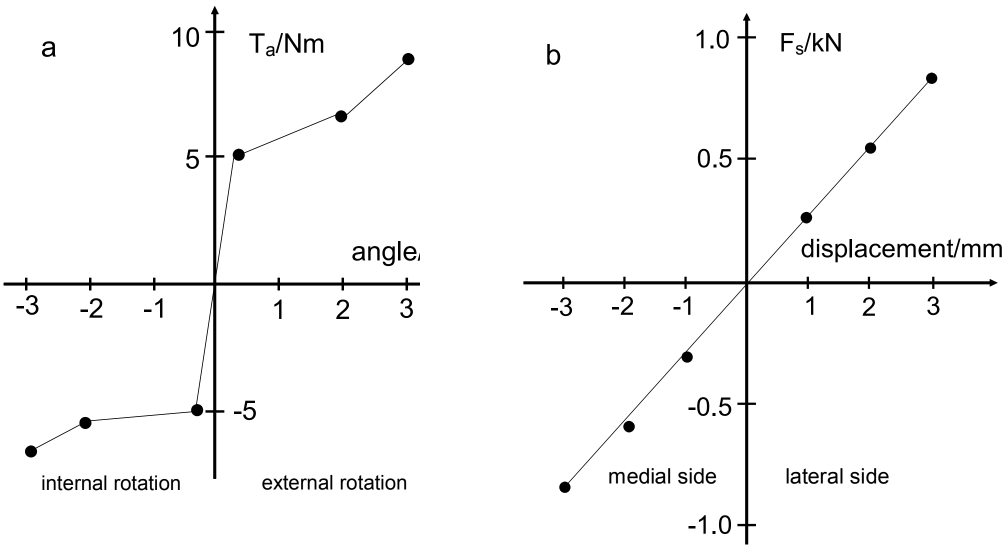

Fig. 7. Luxation characteristics of the AEQUOS-G1 under 2.6 kN axial compressive load: a) counteracting torque T_a as the function of axial rotation; b) counteracting shear force as the function of ab/adduction displacement

This constructional provision makes that under a compressive joint force the TKR works like a crane carriage. But it allows a certain play in small axial rotations and valgus/varus translations especially under external side-impact loads. Under the joint load these kinematical degrees of freedom are self-stabilising. In figure 7, the respective stabilising characteristics (luxation characteristics of AEQUOS-G1) are shown which were measured in vitro by test equipments[2].

[2] The measurements were performed by the IMA firm (Dresden).

3. Tribological testing

Since the decisive criteria for medical approval are the wear tests according to ISO standard draft 14243 [2], we had put in charge the IMA test firm in Dresden to carry out the specified tests for our novel TKR (AEQUOS-G1). The results we have already reported at DAS 2003 [13] show that the wear rates are lower than in conventional TKR possessing the same lubricate pairing.

4. Implantation procedure

An important detail should be mentioned. The tibial slope is integrated in the design of the tibial PE-inlay. Therefore the metal plate, which grasps the PE-inlay (figure 4), has to be implanted perpendicularly to the long axis of the tibia (figure 8).

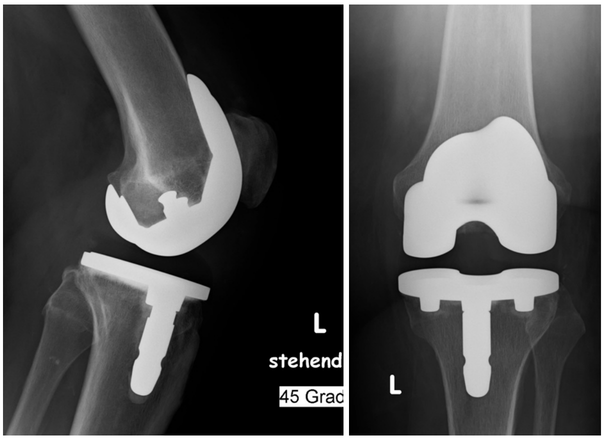

Fig. 8. X-ray photographs of an AEQUOS-G1 three months after surgery.
Note: the plug is parallel to the tibia long axis

5. In vivo evaluation of the rolling back mechanism

3 months after implantation of an AEQUOS-G1, routine checks were taken by lateral X-ray photographs. The patients were asked to bend their treated knee up to about 45°-flexion under the load of body weight. In extension and in flexion, a picture was taken in each case. The position of the patella and therewith the patella tendon angle could be easily and precisely located in any case (figure 9).

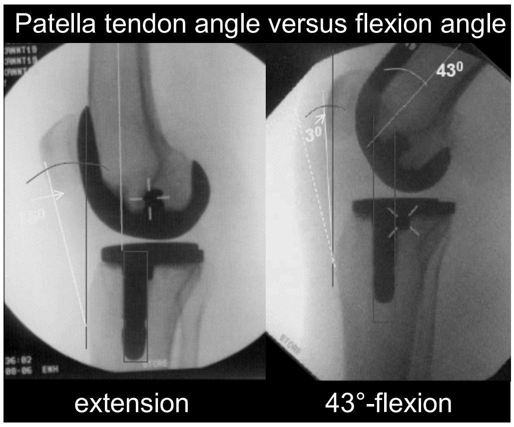

Fig. 9. Measurement of the patella tendon angle as the function of the flexion angle. Superimposition of both pictures yielded: the change of the patella tendon angle $\Delta\beta = 12°$ during 43° flexion

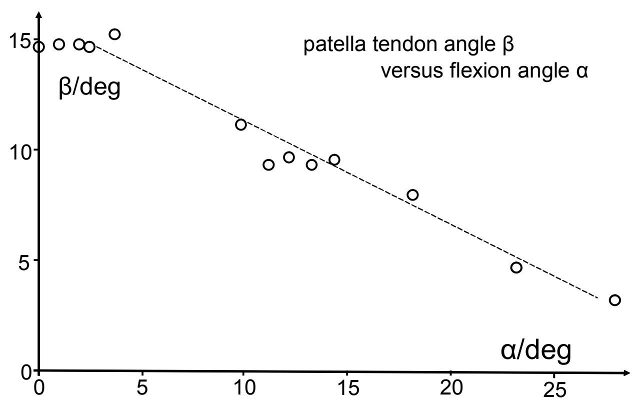

Fig. 10. Patella tendon angle versus flexion angle measured by X-ray fluoroscopy.
This single case shows the ratio of $\Delta\beta/\Delta\alpha = 0.45$

The reference was given by the position of the plug of the metallic component of the tibial implant. By that the alteration of the patellar tendon angle β could be read by superimposing the two pictures by

Table 3. Measurements of the rollback. The rollback of the AEQUOS-G1 prosthesis was close to the natural knee

	Reference	$\Delta\beta/\Delta\alpha$
Conventional TKR	PANDIT [1]	0 …< 0.08
Natural knee	PANDIT [1]	0.3
AEQUOS, single case (fig. 9)	our own measurement	0.28
AEQUOS, 10 cases	our own measurements	0.25 (SD: 0.11)

means of the plug. Ten patients were examined in this way. Results are summarized in table 3.

In one patient provided with AEQOUS-G1, the swiveling of patella tendon backwards could continuously be checked by X-ray fluoroscopy (figure 10).

6. Discussion

The AEQUOS-G1 TKR has a unique design which differs from the other TKRs available on the market. Due to special shaping its functional properties are closer to those of the natural knee in the following aspects:

1. *Kinematics:* the rolling/gliding characteristic makes rolling possible in the stance phase and thus static and gliding friction avoidable in the case of highly compressive joint loads.

2. *Wear and tear:* despite the convex shaping of the lateral tibial compartment the wear of the PE-inlay was small [13] because of the rolling. Additionally tests in our lab showed that under rolling the occurrence of tears and lamellation of the polyethylene were less probable than under sliding.

3. *Stability:* the AEQUOS-TKR gives leeway to ab-/adduction and axial rotation in the cases of external impacts which unavoidably arise during running and walking on unpaved ways. By this the impacts of shear loads are decreased, especially at the interface between tibial implant and tibial bone. After easing off the impacts the prosthesis takes its normal position.

4. *Loading the patellofemoral joint* is decreased by the pronounced rollback of the prosthesis. Therefore we have the hope that for AEQUOS-TKR the appearance of the well-known anterior knee pain will be unlikely as it is in the natural knee.

5. *Ligament balancing:* like the natural knee this prosthesis needs a careful balancing of the ligaments. Therefore, as is a matter of previous experience with the AEQUOS-G1, the posterior cruciate ligament should not be excised.

7. Conclusion

By the development and the successful clinical use of the AEQUOS-G1 a novel type of TKR is available whose mechanical properties are close to those of the natural human knee.

References

[1] PANDIT H., WARD T., HOLLINGHURT D., BEARD D.J., GILL H.S., THOMAS N.P., MURRAY D.W., *Influence of surface geometry and the cam-post mechanism on the kinematics of total knee replacement*, J. Bone and Joint Surgery, 2005, 87B, 940–5.

[2] WALKER P.S., BLUNN G.W., PERRY J.P., BELL C.J., SATHASIVAM Sh., ANDRIACCHI T.P., PAUL J.P., HAIDER H., CAMPBELL P.A., *Methodology for long-term wear testing of total knee replacements*, Clin. Orthop. Rel. Res., 2000, 372, 290–301.

[3] WEBER W., WEBER F., *Mechanik der menschlichen Gehwerkzeuge*, Dietrichsche Buchhandlung, Göttingen, 1836.

[4] FISCHER O., *Kinematik organischer Gelenke*, Vieweg-Verlag, Braunschweig, 1907.

[5] LANGER K., *Das Kniegelenk des Menschen*, Sitzungsberichte der Akademie der Wissenschaften in Wien, Mathematisch-naturwissenschaftliche Klasse, 1858, Vol. 2 and 3.

[6] BRAUNE W., FISCHER O., *Bewegungen des Kniegelenkes, nach einer neuen Methode am lebenden Menschen gemessen*, S. Hirtzel, Leipzig, 1891.

[7] Zuppinger H., *Die active Flexion im unbelasteten Kniegelenk*, Züricher Habil Schr Bergmann Verlag, Wiesbaden, 1904.

[8] WALKER P.S., ROVICK J.S., ROBERTSON D.D., *The effects of knee brace hinge design and placement on joint mechanics*, J. Biomechanics, 1988, 21, 965–74.

[9] WISMANS J., VELDPAUS F., JANSSEN J., HUSON A., STRUBEN P., *A three-dimensional mathematical model of the knee-joint*, J. Biomechanics, 1980, 13, 677–85.

[10] LI G., DEFRATE L.E., PARK S.E., GILL T.J., RUBASH H.F., *In vivo articular cartilage contact kinematics of the knee. An investigation using Dual-Orthogonal Fluoroscopy and Magnetic Resonance Image-Based Computer Models*, The American Journal of Sports Medicine, 2005, 33, 102–107.

[11] PINSKEROVA V., IWAKI A., FREEMAN M.A.R., *The shapes and relative movements of the femur and tibia in the unloaded cadaveric knee: A study using MRI as an anatomical tool*, [in:] *Surgery of the knee*, Third edition, edited by Insall J.N. and Scott R., W.B. Saunders, Philadelphia, 2001.

[12] PINSKEROVA V., JOHAL P., NAKAGAWA S., SOSNA A., WILLIAMS A., GEDROYC W., FREEMAN M.A.R., *Does the femur roll-back with flexion*? J. Bone and Joint Surgery, 2004, 86-B, 925–931.

[13] KUBEIN-MEESENBURG D., ABICHT Ch., THOMSEN M., DATHE H., ADAM P., FANGHÄNEL J., NÄGERL H., *A functional knee-endoprosthesis with rolling articular surfaces during the stance phase*, 20th Danubia-Adria Symposium on Experimental Methods in Solid Mechanics, Györ 24, 27 September 2004.