XVI IMEKO World Congress
Measurement - Supports Science - Improves Technology - Protects Environment ... and Provides Employment - Now and in the Future
Vienna, AUSTRIA, 2000, September 25-28

DYNAMOMETER FOR THE MEASUREMENT OF TORQUES ON HUMAN JOINTS

T. Angeli
Institute for Machine Elements and Machine Design
Vienna University of Technology, A-1060 Vienna, Austria

Abstract: In the field of biomechanics, in sport as well as in rehabilitation, it is very important to deduce the muscle-force knowing the torque of the joints. So called knee-dynamometers are used for this purpose. These dynamometers must follow given velocity-courses of the measuring lever regardless of the torque to be measured. Up to now there are only dynamometers available for measuring static torques and torques with constant velocity. However, these dynamometers can only measure the torque of isometric contractions and extensions or flexions of the knee joint with constant angular velocity. Therefore the idea was born to design and construct a prototype which is able to measure the torque during acceleration-courses for different human joints. At the same time a new measuring method would ensure correct results by considering gravity or inertia of masses during acceleration. It became evident that the force-development of the physiologically contracted muscle differs to that of the electrically stimulated muscle.

Keywords: force/torque velocity relationship, functional electrical stimulation, Human Functions Measurement

1 INTRODUCTION

According to a publication in the European Journal of Applied Physiology neglecting gravitational forces leads to deviations of 26 % to 43 % when measuring extensions and 55 % to 510 % when measuring flexions [16]. The huge deviations during flexion can be explained as follows: The total mechanical work was measured, which means the sum of the muscle work, the potential energy, friction loss, passive muscle force, etc. In case of flexion of the knee joint, the part of potential energy of the measured total work was now extremely large (especially with one person: 510 % !). The mistake was computed when the difference of the measured total work and of the effective muscle work was related to effective muscle work. This is especially evident with muscle-atrophic and/or paralytic patients.

We also intend to use the dynamometer for investigation of the simulation parameters for **F**unctional **E**lectrical **S**timulation (FES). In the case of paraplegic test persons whose muscles are activated by FES, the applied forces decrease very fast due to fatigue and the residual moments are only about 4 Nm. Therefore it is especially important to avoid the mistakes mentioned above.

2 MATERIAL AND METHODS

2.1 Construction of the test bed

The test bed (Figure 1) basically consists of a chair and a measuring arm which both are mounted on a frame. The chair on which the test person is seated (Figure 2) can be moved vertically and the backrest can be moved horizontally. This is necessary in order to set the rotation axle of the

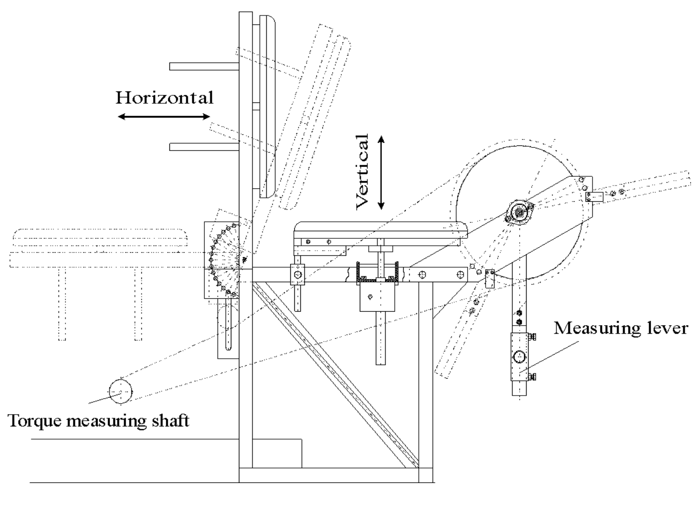

Figure 1: Scheme of the test bed

measuring arm according to the simplified joint-kinematics of the knee. Physiologically the movement in the knee joint is a combination of flexion/extension, translation and internal/external rotation. To consider all parameters of this complex movement, the chair would have to be movable. In the test bed however the movement of the knee joint is reduced to a pin joint (flexion/extension). The backrest can be inclined and fixed in various angles. These adjustments are important because they consider the influence of the two-joint muscles (extension: musculus rectus femoris; flexion: hamstrings and musculus gastocnemius) on the joint torque. Thus the force development and/or the torque development during knee extension changes in the case of specific knee angles for different hip angles (because the moves over the knee and hip joint).

Additionally joint torques of hip, elbow and shoulder joints can be measured by setting the backrest horizontally and prone position of the testperson. A chain transmits the applied torque at the lever to a torque measuring shaft (T 32 FN from Hottinger Baldwin Meßtechnik) which is coupled with a gear motor (AC - Servomotor MAC 71 from Indramat).

Figure 2: Photo of the test bed with test person

2.2 Electronic Control

A servo motor is controlled by a personal computer which also reads the data. The motor is coupled with a torque measuring shaft, which is connected to the measuring arm by a cog belt. To avoid injuries the measuring lever is controlled by an inductive proximity switch and an inductive limit switch which turns the servo motor off. An emergency stop button is also installed. Additionally a mechanical stop is positioned behind the electric ones. As soon as the mechanical limit is reached the cog belt on the cog belt disk will slip. These features guarantee the safety of the testperson in case of power failure or computer-malfunction.

3 MEASUREMENTS

The knee-dynamometer's measurement during contraction of the knee extendsor (musculus quadriceps) are shown in figure 3. First the test person (in figure 3, a sportsman) must warm up. Then the test person puts on the comfortable special shoe which is strapped to the measuring lever (Figure 2). Two measuring steps follow: first the test person is supposed to follow the pre-set velocity process passively i.e. without muscle contraction. The values determined in this case correspond to the passive knee torque in figure 3. The measured knee-torque includes gravity forces of the lower extremity (plus shoe, measuring arm), passive muscle-forces and loss of power due to friction. Secondly the sportsman is requested to extend the knee joint with maximum force. The musculus quadriceps is entirely contracted at a fixed knee flexion angle of 110° (0° knee flexion angle equals fully extended lower extremity). This isometric contraction is shown as the measured knee torque between 0 ms and 1800 ms. Between 1800 ms and 5600 ms the test person performs a concentric contraction which is followed by an eccentric contraction. This steps are done at 15°/s, 30°/s, 45°/s, 60°/s and 90°/s. The test person tries to accelerate the measuring arm during the concentric movement and tries to resist the eccentric movement. In the second diagram of figure 3 the knee angle course over time is also indicated in [ms]. The active knee torque is calculated as the measured knee torque (step two) minus the passive knee torque (step one) and is also represented in figure 3.

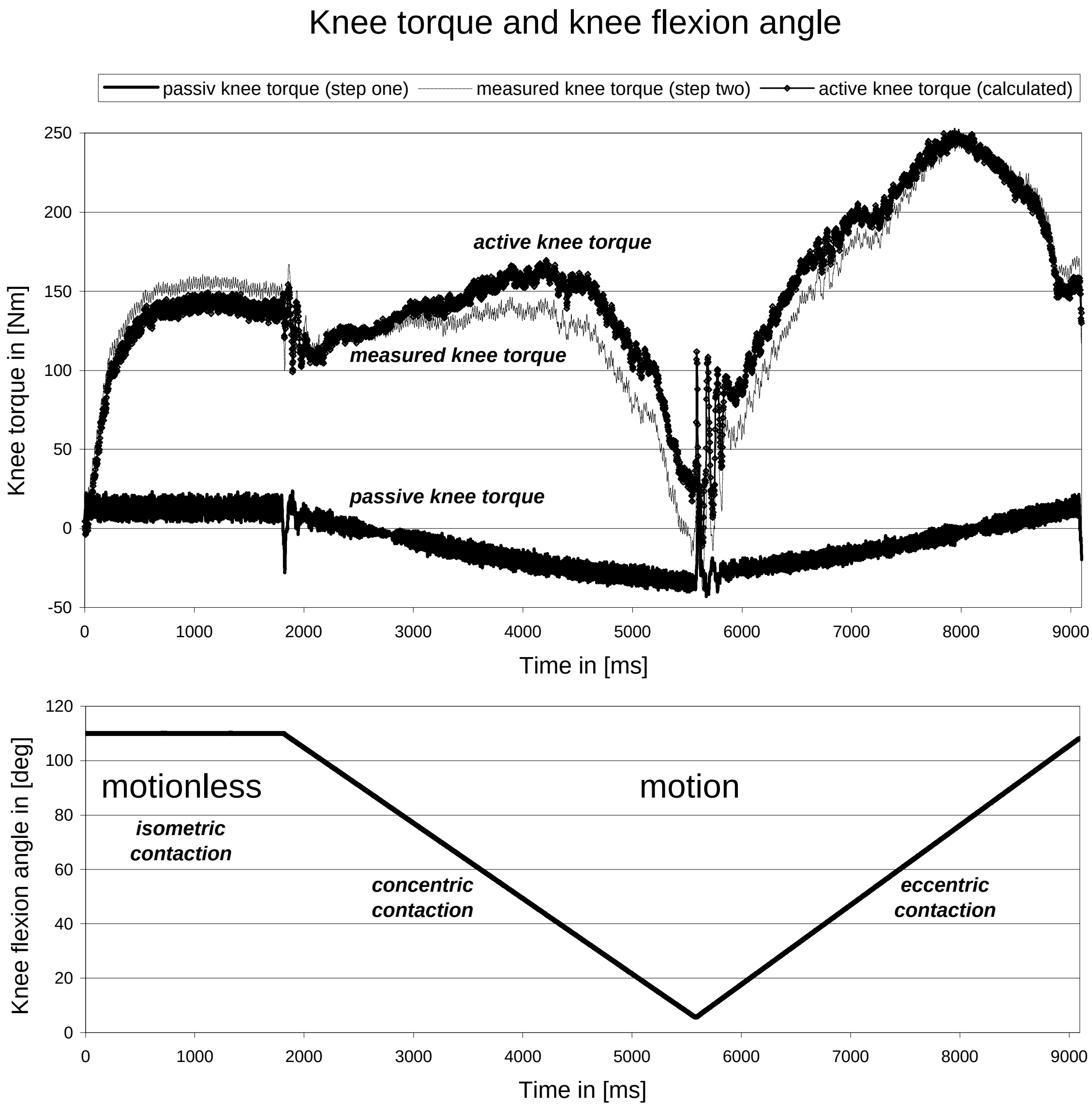

Figure 3: Results of knee torque measurements and knee flexion angle of one measurement cycle by physiological contraction (motionless and at 30°/s). Measurements also done at 15°/s, 30°/s, 45°/s, 60°/s and 90°/s.

The active knee torque indicates that a higher force can be developed during an eccentric movement compared to a concentric movement. The isometric muscle force is higher then the concentric muscle force at the same angle, which can be seen in figure 3 (change from isometric to concentric contraction at 1800 ms: decline of force at the beginning of the movement).

The same measuring-cycle was done by stimulating the muscles with surface-electrodes and the stimulation-unit Compex Sport-p (shown in figure 4, diagram in figure 5). This was done to measure the torque in relation to the knee angle when using FES. The results will be used to improve a movement-optimising simulation program for paraplegic patients stimulated by FES. Reduced torque not only could be observe, but also a different torque in relation to the knee-angle. When stimulated by FES maximum knee torque was measured at smaller knee flexion angles compared to physiological contraction. This difference will be taken into consideration in the FES movement-simulation.

Figure 4: Surface-electrodes and stimulation unit Compex Sport-p

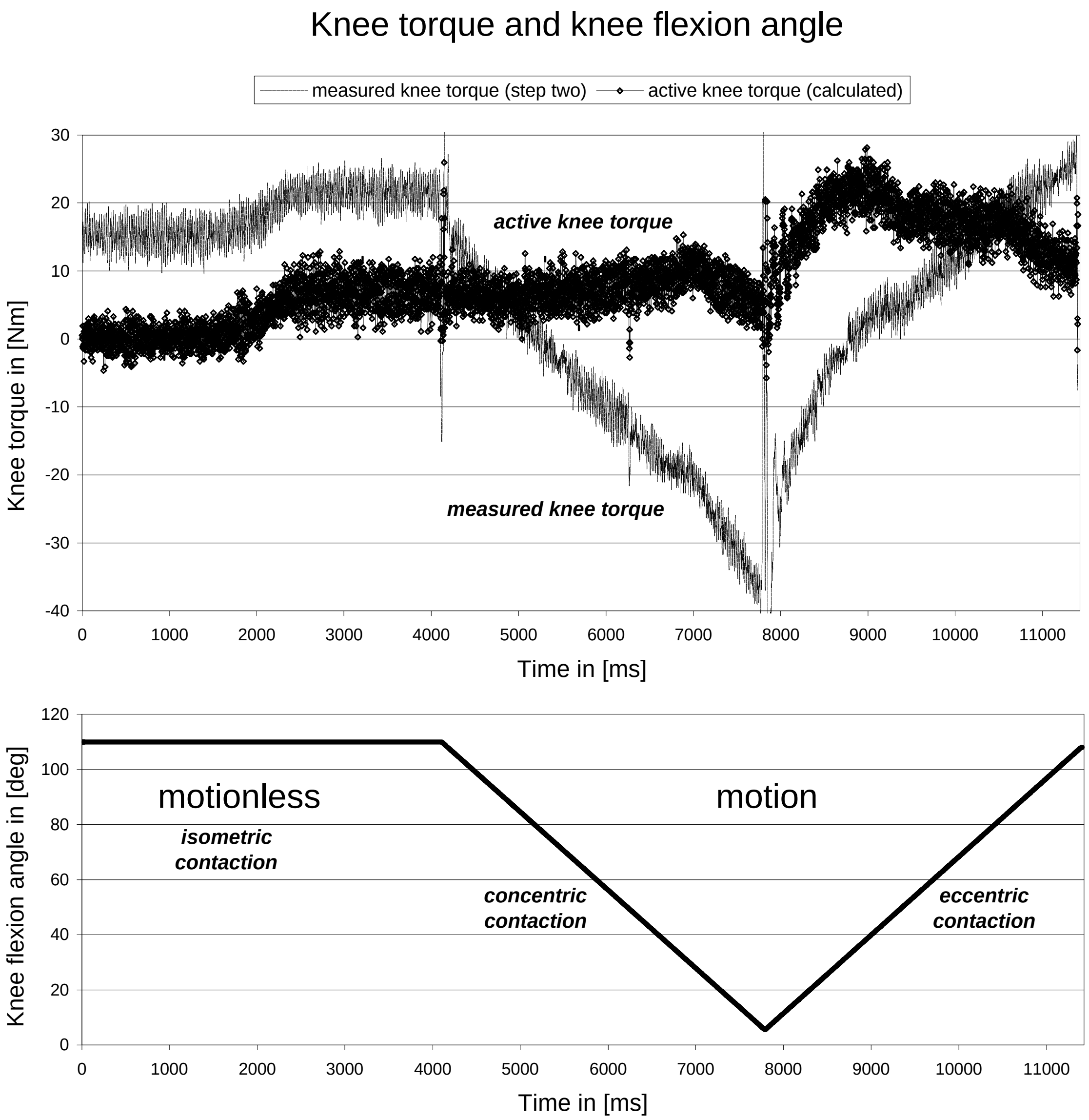

Figure 5: Results of knee torque measurements and knee flexion angle of one measurement cycle by means of FES (motionless and at 30°/s). The passive knee torque is not shown for reasons of clearness. Measurements also done at 15°/s, 30°/s, 45°/s, 60°/s and 90°/s.

To calculate muscles' force, the measured torque must be divided by the muscle moment arm at the corresponding knee flexion angle. The effective moment arm of the musculus quadriceps at the knee joint is defind as the average of Chow's et al. [3] and Pawlik's [12] results in a knee angle between 0° and 120°. From 120° and upwards the moment arm of the quadriceps is constant (Figure 6). Pawlik [12] took the average of Andrews [1], Draganich et al. [4], Grood et al. [5], Hoy et al. [8], Redfield and Hull [13], Spoor and van Leeuwen [15], Yamaguchi and Zajac [17] and Yoschihuku and Herzog [18]. Chow et al. [3] took the average of Bandi [2], Grood et al. [5], Haxton [6], Herzog [7], Kaufer [9], Lindahl and Movin [10], Nisell et al. [11] and Smidt [14].

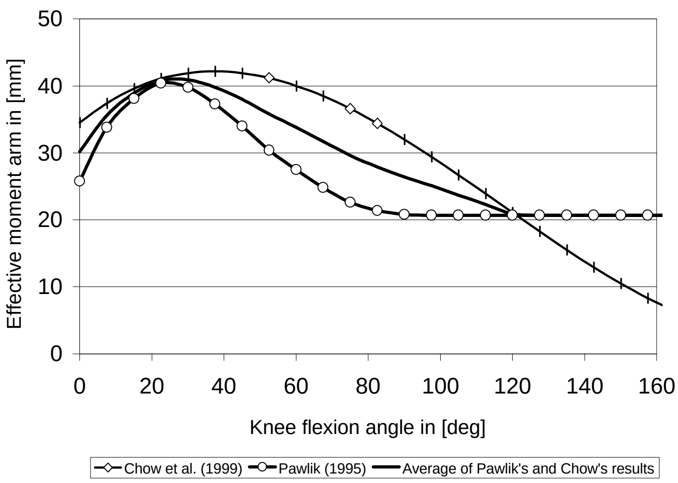

Figure 6: Comparison of effective moment arms of the quadriceps force about the knee centre

The force in relation to the knee flexion angle can be seen in figure 7 and 8.

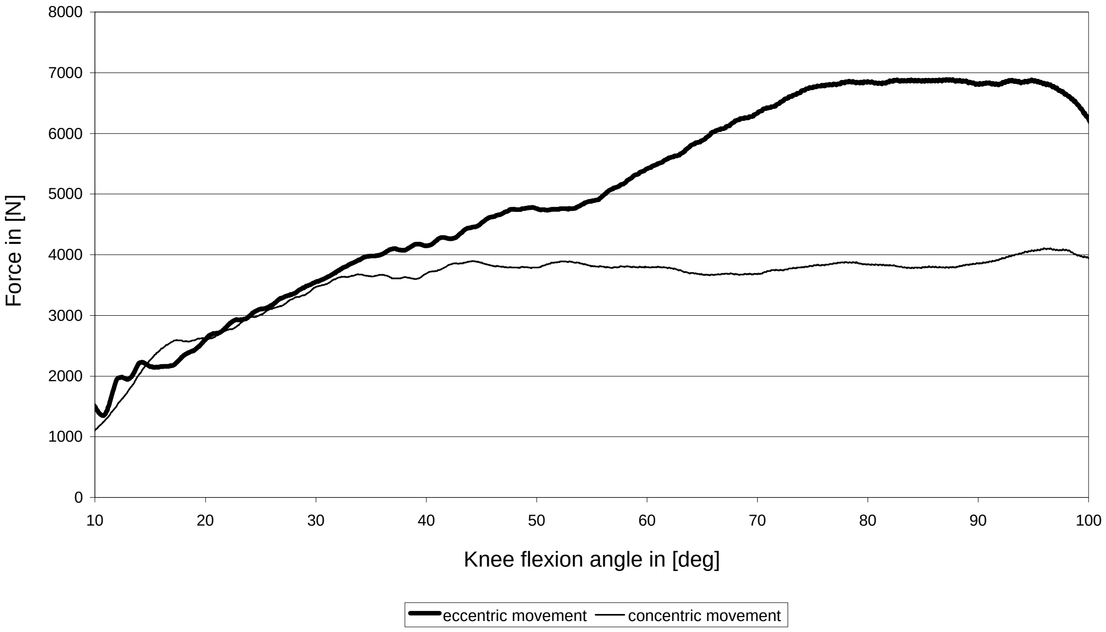

Figure 7: Muscle force physiologically contracted

Figure 8: Muscle force using FES

The comparison of figure 7 and 8 shows a ten times higher muscle force when contracted physiologically than when using FES. This is also because we do not want to cause pain (electrical shock and burns) to the testpersons. After being tested more often higher muscle forces can be reached.

The forces obtained by means of physiological stimulation, both concentric and eccentric muscle force are nearly identical up to a knee flexion angle of 35°. When contracted concentrically the muscle force is constant (approx. 4000 N) at knee flexion angles of 35° and higher; when contracted eccentrically the force increases until the knee flexion angle reaches 75° (figure 7). Another interesting fact is that when stimulated with FES the muscle force reached when moved eccentrically is always clearly higher than when moved concentrically (figure 8).

In conclusion the measurements show that stimulating muscles by means of FES must seriously consider the effect of the antagonistical muscle (eccentrically moving muscle) to avoid cocontraction (no movement), by means of intelligent stimulation timing and to avoid too large electrical field.

4 DISCUSSION

The measurements show that in our case of physiologically contracted muscles the influence of gravity is not as strong as Winter [16] describes. At a knee angle at 20° the active knee torque is 5 Nm by using FES but the measured knee torque is less than –30 Nm (figure 8). When not considering the passive torque (potential energy, friction loss, passive muscle force, etc.) the deviation is more than Winter [16] has describes (more than 700 %). These measurements are taken by using FES on physiologically intact muscles. It is impossible for testpersons to suppress their own nervous impulses totally. To obtain more realistic results we are planning to repeat these tests on paraplegic testpersons

REFERENCES

[1] Andrews, J.G. (1987). The functional roles of the hamstrings and quadriceps during cycling: Lombard's paradox revisited. *J. Biomechanics,* **20**, 565-575.
[2] Bandi, W. (1972). Chondromalacia patelle und femoro-patellare arthose. *Helvetica Chirurgica Acta*, **11** (Supplement), 1-70.
[3] Chow J.W., Darling W.G. and Ehrhardt J.C. (1999). Determining the Force-Length-Velocity Relations of the Quadriceps Muscles: II. Maximum Muscle Stress. *J. of Appl. Biomech.*, **15**, 191-199.
[4] Draganich, L.F., Andriacchi, T.P. and Anderson, G.B.J. (1987). Interaction between intrinsic knee mechanics and the knee extensor mechanism. *J. orth. Res.,* **5**, 539-547.
[5] Grood, E.S., Suntay, W.J., Noyes, F.R. and Butler, D.L. (1984). Biomechanics of the knee extension exercise, effect of cutting the anterior cruciate ligament. *J. Bone Jt,Surg.,* **66A**, 725-735.
[6] Haxton, H.A. (1945). The function of the patella and the effects of its excision. *Surgery, Gynecology and Obststrics*, **80**, 389-395.
[7] Herzog, W.H. (1985). Individual muscle force prediction in athletic movements. Ph.D. thesis, Univ. of Iowa, Iowa City.
[8] Hoy, M.G., Zajac, F.E. and Gordon, M.E. (1990). A musculosceletal model of human lower extremity: The effect of muscle, tendon and moment-arm on the moment-angle relationship of musculotendon actuators at the hip, knee and ankle. *J. Biomechanics,* **23**, 157-169.
[9] Kaufer, H. (1971). Mechanical function of the patella. *J. of Bone and Joint Sugery,* **53-A**, 1551-1560.
[10] Lindahl, O. and Movin, A. (1967). The mechanics of extension of the knee-joint. *Acta Orthopaedic Scandinavica,* **38**, 226-234.
[11] Nisell, R., Nemrth, G. and Ohlsen, H. (1986). Joint forces in extension of the knee – Analysis of a mechanical model. *Acta Orthopaedic Scandinavica,* **57**, 41-46.
[12] Pawlik R. (1995). Biomechanik des Radfahrens. Untersuchung der maximalen menschlichen Leistungsabgabe. Ph.D. thesis, Vienna Univ. of Techn., Österr. Kunst und Kulturverlag, Vienna.
[13] Redfield, R. and Hull, M.L. (1986). Prediction of pedal forces in bicycling using optimization methods. *J. Biomechanics,* **19**, 523-540.
[14] Smidt, G.L. (1973). Biomechanical analysis of knee flexion and extension. *J. Biomechanics,* **30**, 79-92.
[15] Spoor, C.W. and van Leeuwen, J.L. (1992). Knee muscles moment arms from MRI and tendon travel. *J. Biomechanics,* **25**, 201-206.
[16] Winter D. A., Wells R. P. and Orr G. W. (1981): Errors in the Use of Isokinetic Dynamometers. European Journal of Applied Physiology. **46**: 397-408.
[17] Yamaguchi, G.T. and Zajac, F.E. (1989). A planar model of the knee joint to characterize the knee extensor mechanism. *J. Biomechanics,* **22**, 1-10.
[18] Yoschihuku, Y. and Herzog, W. (1990). Optimal design parameters of the bicycle-rider system for maximal power output. *J. Biomechanics,* **23**, 1069-1079.

AUTHOR: Dr. Thomas Angeli, Institute for Machine Elements and Machine Design, Vienna University of Technology, Getreidemarkt 9/306, A-1060 Vienna, Austria, Phone Int. +43 1 58801 30613, Fax Int. +43 1 58801 30699, E-mail : thomas.angeli@tuwien.ac.at