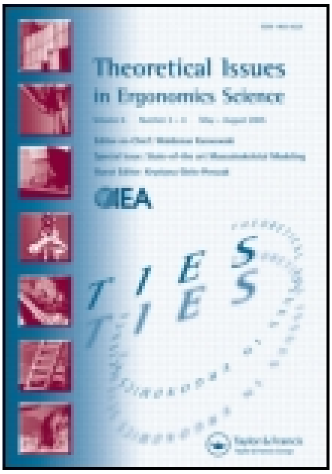

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The history of musculoskeletal modelling in human gait

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The history of musculoskeletal modelling in human gait

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This paper presents a brief resumé of the historical development of instrumented gait analysis leading to analysis of force transmitted in the human body by minimisation techniques and by implanted transducers.

Keywords: Musculoskeletal modelling; Gait analysis; Human locomotion

1. History

The early investigators into the mechanics of body movement, Leonardo da Vinci (c1500) and Borelli (1679) among others, based their work on a knowledge of anatomy and observation of the way in which the human body was used in various activities. They did not have any equipment which would allow them to measure the patterns of movement or the forces developed in movement of the human. The science of movement analysis of the human progressed with the work of the Weber brothers (1836) who used a telescope with a calibrated graticule to measure as best as they could the movement of specific anatomical points on the human during walking. They produced some realistic diagrams of the anatomical configuration of the walking human although close inspection of them suggests that they showed excessive amounts of knee flexion during the stance phase and they postulated various theories including the 'pendulum' action of the leg in the swing phase. Real progress was made however by Marey (1873) and co-workers who developed a pneumatic system of measurement of movement having a pressure sensor under the foot and a revolving recorder carried in the hand of the test subject. This allowed interpretation of the temporal factors of gait and was applied also to movements of the body segments. Marey also used a single plate camera with an automated shutter to record on a single negative the successive positions of reflective markers over the locations of the centres of the joints of the leg. This produced two dimensional illustrations of the locomotion of the human as a succession of 'stick' diagrams. The quantitative measurement in three dimensions of the movement of the human body was greatly advanced by the work of Muybridge (1882) who published a wide series of pictures of the frontal and sagittal plane movements of test subjects undertaking a range of activities. These were obtained by a series of cameras set up together and connected so that their shutters were opened consecutively. Many investigators of human gait conduct tests in which the test subject's locomotion is undertaken 'bare-foot'. Arguments can be made for and against this although

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in the Western world it is unusual to see many activities of the normal day being conducted in this way. Muybridge carried this idiosyncrasy even further in that his test subjects wore no apparel whatsoever!

The early investigations into human movement were concerned principally with the kinematics and temporal factors of locomotion. However, if the measurements are sufficiently precise it is possible to undertake double differentiation to obtain the relevant linear and angular accelerations. Braune and Fischer (1890) undertook a frequently cited investigation into the mass properties of the segments of the human body, in particular the position of the centre of mass as a proportion of the length of the body segment from the proximal joint and the second moment of mass about a set of perpendicular axes through the centre of mass of the segment. The centre of mass was presumed to be on the line joining the centres of the proximal and distal articulations of the segment and the second moment of mass was taken to be the same in the sagittal and frontal planes. They also obtained the second moment of mass about a longitudinal axis through the centre of mass but the differences quoted between right leg and left leg suggest that those may be approximate. Following this work Fischer (1891–1904) undertook a series of investigations funded by the German Government and intended to provide information on the best method of load carriage by the German soldier when marching. In a test laboratory, flashing lights were fixed to points relating to the axes of the joints of the human body including arms, head, trunk and legs. Locomotion took place in a darkened room with the test subject viewed by 4 single plate cameras viewing the test subject along axes $\pm 30^\circ$ to the direction of progression from the front and from behind. In an extensive series of calculations, Fischer determined the linear and angular displacements of velocity and accelerations for the centres of mass of the body segments and where possible calculated the inter-segment moments. It was not possible to calculate these for the legs in the period of double support since this involves 6 unknown quantities corresponding to 3 components of force and the moments of these forces about 3 reference axes for each foot—a total of 12 unknowns. The whole body has six equilibrium equations and thus the loading system in double support is statically indeterminate and no values can be calculated from displacements and accelerations alone. The work of Bernstein (1935) in the Soviet Union was more extensive in respect of numbers of test subjects and numbers of tests performed than that of Fischer but being published in the Russian language generally has not been reviewed extensively in the Western literature. The American physiologist Elftman (1938) used a ground to foot force measurement platform where components of ground to foot force were quantified by the deflections of a series of springs. The instrument however did not measure the six quantities necessary for a full description although he did have the advantage of the newly developed cine camera for his locomotion assessments.

A pair of volumes under the title ‘Fundamental studies of human locomotion’ was produced at the Berkeley campus of the University of California in 1947 with the principal editor Professor H. D. Eberhart of the Department of Civil Engineering. This comprises the results of a two-year contract from the United States Health and Veterans Administration. They report studies of normal locomotion with a view to improving the locomotion, the prostheses and the treatment of amputees. They undertook three dimensional movement analysis using three 35 mm cine cameras simultaneously and incorporated glass walkways with a mirror underneath so that a further camera could measure the areas of foot to ground contact. They further developed instrumentation which has provided essential information and has allowed full analysis of the locomotion of the human. Cunningham and Brown (1952) reported separately the design and characteristics of two instruments: one a six quantity force transducer for incorporation in the leg of the prosthesis of an amputee; the other a six quantity ground to foot force transducer—the first instrument allowing continuous recording of the six quantities characterising ground to foot force. These instruments were based on electrical resistance strain gauges and had problems with sensitivity, cross sensitivity, and in the case of the force platform, vibrations at natural frequency. They formed the basis for force platforms as we now know them manufactured with up to date strain gauge technology and transducer design or piezo-electric technology. The Berkeley campus output has included the classic work by Saunders *et al.* (1953) 'The major determinants in normal and pathological gait' together with studies on amputee locomotion and the design of knee mechanisms and feet for prostheses.

The ability to measure ground to foot force during double support was particularly important for further developments in analysis of loading at joints in the leg during locomotion since the maximum values of these forces generally occur during the periods of double support. This group also developed the basic ideas on measurement of myo-electric (EMG) signals reported by Proebster (1928). The use of phasic patterns of EMG in gait analysis was further developed by Joseph and Williams (1957), Close and Todd (1959) and Basmajian (1967).

Berkeley again provided the first major record of inter-segment forces and moments in the leg during locomotion in a paper by Bresler and Frankel (1950). These authors presented curves of variation through the gait cycle of the components of inter-segment force and components of inter-segment moment. It should be noted that inter-segment force is not joint force but corresponds to the summation of the external forces distal to the joint (gravity, inertial and ground to foot) together with the tensions in muscles or ligaments connecting the segments at the joint under consideration. These authors speculated that their data could be used to determine the forces at the joints but thought that the complexity of anatomical function and structure would prevent reasonable calculations from being made. This however has become a study of some complexity as a specialist field in biomechanics. Having an interest solely in joint forces Paul (1967) with the help of surgical colleagues took the view that the action of the 21 muscles at the hip could realistically be represented by six equivalent muscles with two tending to produce flexion, two tending to produce extension, one abduction and the other adduction. It was recognised that there were muscles acting to produce internal and external rotation at the hip but it was also recognised that the muscles of much larger cross section producing moments about the other axes were in general inclined to the axis of the bony segment and offset from it. These therefore also produced inward or outward rotation due to the small angle of inclination of their line of action relative to the axis of the femur. It was considered therefore that load actions corresponding to torsion about the hip/knee axis could contribute to the analysis only approximately and probably inaccurately. At successive instants in the gait cycle a computer programme determined the sense of the flexion/extension and adduction/abduction loading selected and appropriate muscle groups to resist these. The major muscles were modelled into groups, two tending to produce flexion and two tending to produce extension. The two groups for flexion and extension corresponded to muscles having actions at both hip and knee joint and to muscles having an action at the hip joint only. The first solution considered the two joint muscles since their lever arms were greater than those of the one joint muscles and they give a solution for the least possible force which could be transmitted at the hip joint at that instant. A solution was then taken using the one joint muscles and was designated the upper limit of joint force. It was recognised that this might not be the case since there could well be antagonistic activity between the relevant muscles. However, EMG studies on the major muscles indicated that at the instants of highest joint loading there was comparatively little antagonistic muscle action. Morrison (1965) and Poulson (Paul and Poulson 1974) undertook analyses relating to the knee and the hip and Procter (Procter and Paul 1982) analysed the load transmission at the ankle joint using the same techniques. Morrison's work at the knee joint deserves mention because it was realised that the complexity of the knee joint required that the relative position of tibia and femur be taken account of at each instant and that the knee joint comprised three articulating surfaces namely the patella/femoral, and the medial and lateral tibio/femoral compartments. Since interest at that time was primarily in the tibio/femoral joint, the analysis considered the forces in these separately so that small values of adducting or abducting moment could be transmitted by differences in the medial and lateral joint forces without any muscular or ligamentous tension being involved. Higher values of abducting/adducting moment tended to cause the joint to open and this was taken to be resisted by the relevant lateral or medial collateral ligaments. The resultant anterior/posterior force was transmitted either by the posterior or anterior cruciate ligament. The model included the patellar tendon as the structure tending to extend the knee and either the hamstring muscles or gastrocnemius tending to resist extension. EMG studies showed that there was minimal activity in the hamstring muscles in late stance when gastrocnemius was active due to its function at the ankle. A simplified illustration of Morrison's analysis is shown in Paul (1988).

Instrumented measurement of linear and angular displacements was considerably advanced by Furnee (1967) who developed a television/computer system for automatically recording 2D moment and Jarret *et al.* (1976) who developed the system to allow 3D measurement. These opened the door to the many opto-electronic movement analysis systems in today's market place.

Musculoskeletal modelling has developed into greater complexity with antagonistic muscle action being considered and the whole leg from the hip to ankle or even to fore foot being considered with all fifty muscles in each leg having an action. The big problem in such analyses is that at the hip for instance there are twenty-one muscles although during normal function ligaments are not involved and the components of hip joint force total three. Thus, to analyse the whole loading system required determination of twenty-four unknown quantities from the six equilibrium equations. The first study of this type was undertaken by Seireg and Arvikar (1975) and they applied computer technology to minimise certain parameters including joint force, inter-segment force and others to get a solution, although their initial data are not relevant since the loading conditions which they assumed at ground to foot do not correspond to normal locomotion. Other analysts have used solutions based on minimisation of energy consumption, minimisation of the sum of muscle forces, restriction of muscle force to the product of physiological cross sectional area and a specified stress, minimisation of the sum of weighted sums of a function of muscle force, a function of ligament force and a function of joint force. Others have explored muscle physiology in respect of length/tension variation, force/velocity relationship and muscle force/time of contraction relationships together with the use of EMG data. EMG can be used to identify muscles active at a specific instant in the motion cycle but usually surface electrodes are used and therefore the signals obtained from the major muscles near the skin surface are utilised in these analyses. Others produce their results as periods of activity of the muscles during the walking cycle which are compared with the data from EMG as a validation. It should be noted however that there is considerable variability between the phasic patterns of EMG signals in the literature: it appears that these signals can only be used validly if they are obtained from the test subject at the same time as the locomotion analysis is undertaken. There must be considerable doubt about 'validation' from test subjects in one biomechanics laboratory compared with EMG phasic relationships obtained on different subjects in another laboratory. The statement of Collins (1995) that there has been no validation for any of these minimisation procedures, remains unchallenged.

There have been a few investigations where patients having partial or total hip joint replacements have received an implant incorporating a transducer with provision for measurement of joint load. Rydell (1966) reported data from two test subjects, English and Kilvington (1979) reported data not of joint force but on longitudinal force in the neck of the prosthesis and Bergmann *et al.* (1993) reported data from three test subjects with radio telemetry data transmission from the strain gauges on the prosthesis. The data from these indwelling transducers can be compared with the information from analyses but the transducer data cannot differentiate between muscle models of different types which give the same joint force in particular circumstances. Results from other implanted transducers are summarised in Paul (1999). Data have also been obtained from a partial replacement of the shaft of the femur, where the section was necessary to treat a tumour, and a telemetering load measuring transducer was incorporated in the implant (Taylor *et al.* 1988). Extrapolation from the data from this patient allowed assessment of knee joint forces for comparison with models although it would not be expected that following such radical surgery the patient's locomotion characteristics could be comparable to those of patients with total knee replacement.

A major difficulty in comparison of the modelling predictions of joint load and the few measurements made with implanted transducers is the deficiencies in the literature in respect of complete recording of the data. Generally body mass is reported but although walking speed is referred to it is not always quantified. The type and characteristics of footwear is rarely mentioned and very rarely is cadence or stride length quoted. From mean walking velocity, and either cadence or stride length, all temporo/spatial parameters can be determined. However it should be recognised that, in the leg, joint force is primarily affected by stride length since the moment about an axis through the knee or hip joint is a direct function of stride length; also the value of the vertical component of ground to foot force relates directly with stride length. Figure 1 shows the approximate form of the curve of resultant hip joint force to a base of time involving peak values at approximately 12 and 50% of the gait cycle time. Figure 2 shows values for hip joint force obtained in a range of investigations reported in the literature where sufficient information was provided to infer mean walking speed and/or stride length. The squares show calculated values and the triangles measured values from implanted transducers. The regression lines for the calculated and the measured values are shown. The diagrams show that stride length is a more relevant description of the biomechanics of joint loading. Figure 2d highlights the different characteristics of the patients with

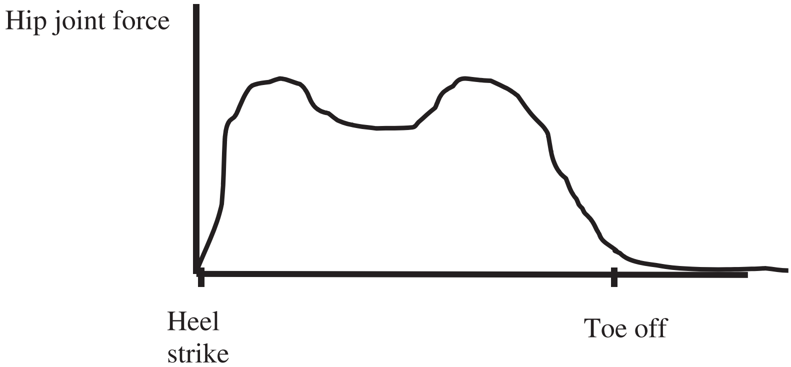

Figure 1. Typical curve of variation of resultant hip joint force with time as a fraction of cycle time.

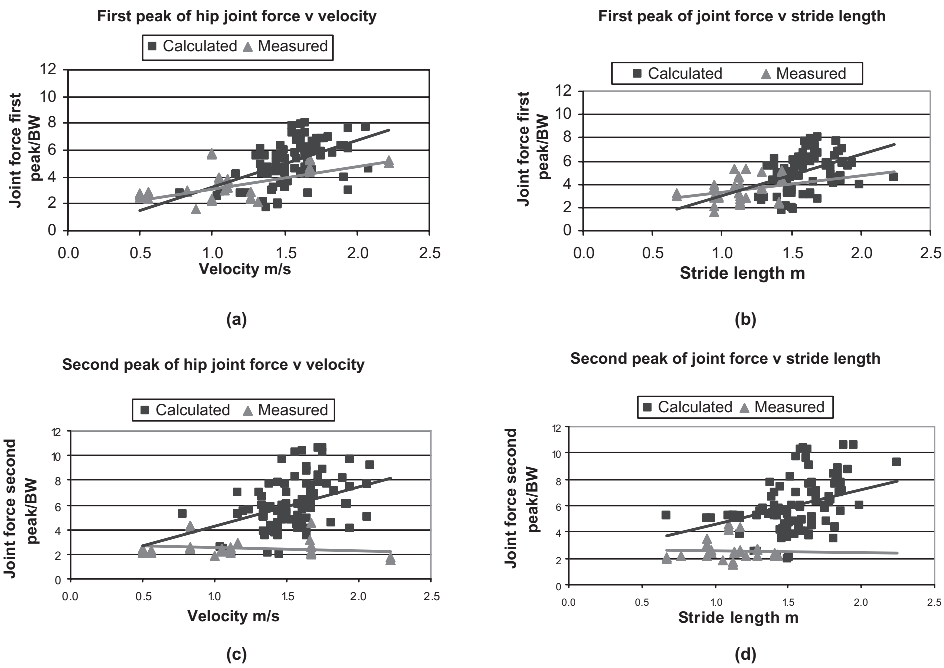

Figure 2. Values of the first and second peaks of the curve of figure 1 from gait laboratory calculations (squares) and from implanted transducers (triangles). The sources of the data are cited† in the reference list. (a) First peak vs. forward speed, (b) first peak vs. stride length, (c) second peak vs. forward speed and (d) second peak vs. stride length. Reproduced with permission from Proceedings of 11th International Conference on Biomedical Engineering, Singapore.

implanted transducers: the second peak of hip joint force plotted to a base of stride length clearly shows that the patients with an implanted transducer had a markedly different gait from the others represented.

An appeal is therefore made to all journal contributors of papers relating to locomotion biomechanics that the essentials of footwear, velocity and stride length together with other parameters are reported in every case.

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†Sources of data in figure 2.

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