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Hip contact forces and gait patterns from routine activities

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Abstract

In vivo loads acting at the hip joint have so far only been measured in few patients and without detailed documentation of gait data. Such information is required to test and improve wear, strength and fixation stability of hip implants. Measurements of hip contact forces with instrumented implants and synchronous analyses of gait patterns and ground reaction forces were performed in four patients during the most frequent activities of daily living. From the individual data sets an average was calculated. The paper focuses on the loading of the femoral implant component but complete data are additionally stored on an associated compact disc. It contains complete gait and hip contact force data as well as calculated muscle activities during walking and stair climbing and the frequencies of daily activities observed in hip patients. The mechanical loading and function of the hip joint and proximal femur is thereby completely documented. The average patient loaded his hip joint with 238% BW (percent of body weight) when walking at about 4 km/h and with slightly less when standing on one leg. This is below the levels previously reported for two other patients (Bergmann et al., Clinical Biomechanics 26 (1993) 969–990). When climbing upstairs the joint contact force is 251% BW which is less than 260% BW when going downstairs. Inwards torsion of the implant is probably critical for the stem fixation. On average it is 23% larger when going upstairs than during normal level walking. The inter- and intra-individual variations during stair climbing are large and the highest torque values are 83% larger than during normal walking. Because the hip joint loading during all other common activities of most hip patients are comparably small (except during stumbling), implants should mainly be tested with loading conditions that mimic walking and stair climbing.

Keywords: Hip joint; Total hip replacement; Force; Measurement; Gait analysis; Telemetry

1. Introduction

Contact forces in the hip joint must be known for tests on strength, fixation, wear and friction of implants, for optimising their design and materials by computer simulation and for giving guidelines to patients and physiotherapists as to which activities should be avoided after a replacement. The movement in the hip joint has to be known when implant wear is tested or the load directions relative to the pelvis are calculated from the forces acting at the femur.

Hip contact forces based on gait analysis data were previously calculated using simplified muscle models and various optimisation methods (Paul, 1967, 1974, 1975; Crowninshield et al., 1978a, b; Röhrle et al., 1984; Brand et al., 1994; Pedersen et al., 1987). Most studies were restricted to walking or stair climbing. Typically the calculations delivered higher hip joint forces than those measured by other groups. Only Brand et al. (1994) compared calculated and measured data which were obtained, however, at different times.

Hip contact forces measured in vivo with instrumented implants were first obtained by Rydell (1966a, b); English (1977, 1978) and English and Kilvington (1979), and more data is available from Davy et al. (1988, 1990) and Kotzar et al. (1988). Loads from patients with tumour implants were published by Taylor et al. (1997, 1998). Hodge et al. (1986) measured the joint pressure in two patients. Most of this literature reports only short-term data from one or two patients when walking or stair climbing. None of them contains detailed gait data.

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We developed two types of instrumented hip implants with telemetric data transmission (Bergmann et al., 1988; Graichen and Bergmann 1991; Graichen et al., 1999). Long-term results from the first two patients were published for a variety of activities (Bergmann et al., 1993, 1994, 1995a, b). In the meantime hip contact forces from seven patients with nine implants were collected up to nine years postoperatively and will be published soon.

The goal of this study was to create an unique data base of hip contact forces and simultaneously measured gait data for future improvements of hip implants. For this purpose measurements were taken in four patients during nine heavy-loading and frequent activities of daily living. A new mathematical averaging procedure was developed to calculate 'typical' results from the data of various trials and patients.

The obtained gait data was used as an input for a musculo-skeletal model to calculate muscle forces (Heller et al., 2001). The measured hip contact forces served to check the validity of calculated results. For walking and stair climbing measured and calculated contact forces agreed fairly well. Their model can therefore be used to investigate clinical problems like muscle deficiencies or operative procedures. Morlock et al. (2001) measured the activity levels of 31 patients with hip implants during day-long sessions. The combination of average activity numbers with the typical hip contact forces and joint movements presented here can serve to test the strength, fixation stability and wear properties of hip implants more realistically than today. Adding the muscle forces of Heller et al. (2001) will make the test conditions for hip implants, femur and pelvis even more realistic. Physiological loading conditions are mandatory if bone remodelling or implant subsidence is investigated (Duda et al., 1998). From the combined data test scenarios of different complexity will be defined soon for simulator tests and computer simulations.

The data volume of measured contact forces and gait data is far too large for inclusion in the text. Therefore, only those hip joint loads are presented which are probably most important for the implant stability. These are the contact force and the torsional moment acting around the stem axis of the prosthesis. Complete data as well as more details about the applied methods are contained on the compact disc 'HIP98'.[2]

[2]The CD 'HIP98' is delivered with this journal. It allows the animated, synchronous display of hip joint loads, gait patterns, muscle forces and patient videos, but also the extraction of all numerical data. It contains detailed descriptions of all measured variables and of methods for transferring data between the different coordinate systems. The mathematical procedure for averaging trials and patients is described there. The data of Morlock et al. (2000) and Heller et al. (2000) are included. The CD is additionally available from the first author.

2. Methods

2.1. Instrumented implants

Two kinds of instrumented total hip implants with telemetric data transmission were used to measure the hip contact forces with an accuracy of 1% at a rate of approximately 200 Hz. The titanium implants had an alumina ceramic head and a polyethylene cup. An implant of type 1 (Bergmann et al., 1988; Graichen and Bergmann, 1991) was cemented in patient IBL, the other three patients got non-cemented prostheses of type 2 (Graichen et al., 1999). The patient images and implant signals from all measurements were stored on video tape for detailed analyses.

The hip contact force with the magnitude F and the components $-F_x, -F_y, -F_z$ was measured in the 'femur coordinate system' x, y, z (Fig. 1). It is transmitted by the acetabular cup to the implant head; the angles of

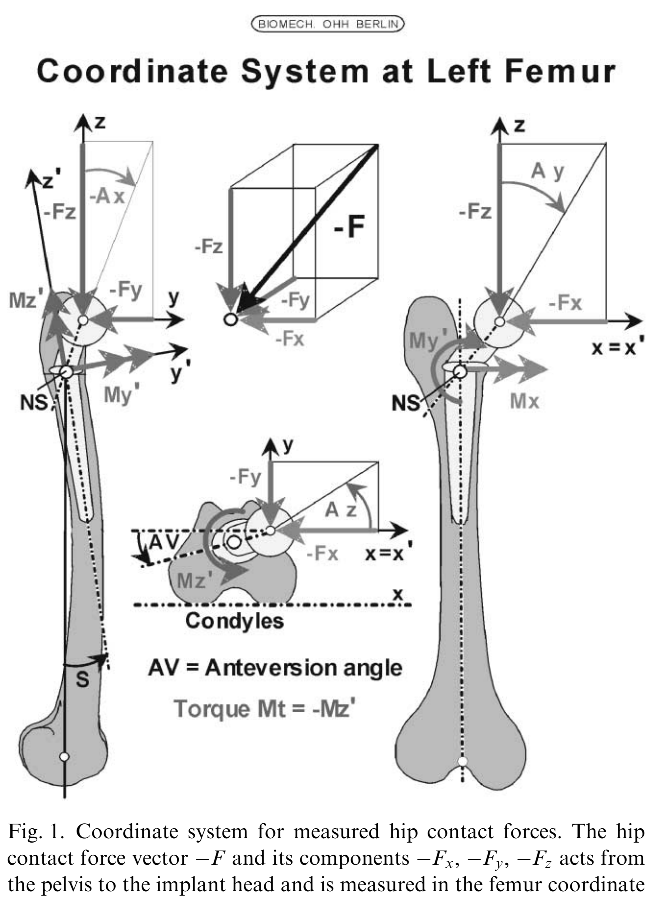


Fig. 1. Coordinate system for measured hip contact forces. The hip contact force vector $-F$ and its components $-F_x, -F_y, -F_z$ acts from the pelvis to the implant head and is measured in the femur coordinate system x, y, z. The magnitude of contact force is denoted as F in the text. The axis z is parallel to the idealised midline of the femur, x is parallel to the dorsal contour of the femoral condyles in the transverse plane. The contact force causes a moment M with the components $M_x, M_{y'}$ and $M_{z'} = -M_t$ at the point NS of the implant. A positive torsional moment M_t rotates the implant head inwards. M is calculated in the implant system x, y', z'. Both systems deviate by the angle S. AV is the anteversion angle of the implant.

inclination of F in three planes are denoted as A_x, A_y, A_z. The force F causes an 'implant moment' M around the intersection point NS of shaft and neck axes of the implant. The moment components $M_x, M_{y'}, M_{z'}$ act clockwise around the axes x, y', z' of the 'implant system' which is rotated by the angle S relative to the femur coordinate system. Important for the implant fixation is the torsional moment $M_t = -M_{z'}$ in the transverse plane which rotates the implant inwards around the shaft axis. The other two components of M are of minor importance; they depend on the definition of the point NS around which M is calculated. The coordinate systems and measured variables are described in more detail in Bergmann et al. (1993) and on the CD.

2.2. Patients

Four patients agreed to participate in this study (Table 1) and to publication of their images and names. They were 51–76 yr old and obtained their implants due to coxarthrosis (patients HSR, KWR, IBL) or a femoral head necrosis after a fracture (PFL). Measurements were taken 11–31 months after implantation. Walking behaviour and mobility was good in patients HSR, PFL and KWR. Patient IBL had a slightly unsafe and unsymmetrical gait due to pain in the opposite hip joint which had been replaced about 10 yr earlier. From the average data of the individual patients data for a 'typical patient NPA' was calculated. NPA is representative for the investigated group of individuals.

Table 1
Personal data and anatomical parameters of patients

Patient	HSR	PFL	KWR	IBL
Gender	Male	Male	Male	Female
Age at implantation (years)	55	51	61	76
Operated joint	Right	Left	Right	Left
Measurement (months postoperatively)	14	11	12	31
Weights at measurement (N)				
Total body (BW)	860	980	702	800
Thigh	75.7	80.2	62.2	98.4
Shank	39.1	53.9	36.5	43.0
Foot	9.2	12.3	9.9	7.2
Lengths (cm)				
Body height	174	175.0	165.0	170
Thigh	43.3	41.0	39.3	47.5
Shank	38.1	41.0	40.0	40.9
Foot	30.0	27.5	29.0	26
Angles (degree)				
AV = Anteversion	4	23	–2	14
S = Femur shaft—implant shaft	10	7	9	9

2.3. Activities

Nine different activities were investigated which are assumed to cause high hip joint loads and occur frequently in daily living (Table 2). Most exercises were performed 4–6 times (trials) by each patient (Table 3). Verbal advice was given to the patients about the speed of walking. This led to only slightly different speeds for slow and fast walking. Due to data errors some trials could not be evaluated. Several activities were not measured in IBL. If KWR or IBL had obvious difficulties to perform an exercise in a normal way, these activities were not used for calculating the typical patient NPA.

For walking and stair climbing the start of the activity cycles was determined by the instants of foot contact. The start and end of cycles for the other activities were chosen from the hip contact forces and synchronous videos.

2.4. Gait analysis

Gait analysis plus contact forces delivered 209 measured and calculated variables (Table 4 and CD). A Vicon system with six cameras and a sampling rate of 50 Hz was used to measure the positions of body markers which were smoothed with 5th order splines. Two Kistler plates measured the ground reaction forces. All data from gait analysis and the readings from the instrumented implants were synchronised using a common marker signal. An interpolation of the

Table 2
Investigated activities

Activity	Abbreviation	Description
Slow walking	WS	Walking at slow speed on level ground, average speed of all patients $v = 3.5$ km/h (0.98 m/s).
Normal walking	WN	Walking at normal speed on level ground, average speed of all patients $v = 3.9$ km/h (1.09 m/s).
Fast walking	WF	Walking at fast speed on level ground, average speed of all patients $v = 5.3$ km/h (1.46 m/s).
Up stairs	SU	Walking upstairs, stair height 17 cm, no support at hand rail.
Down stairs	SD	Walking downstairs, stair height 17 cm, no support at hand rail.
Standing up	CU	Standing up, chair height 50 cm, arms hold at chest height.
Sitting down	CD	Sitting down, chair height 50 cm, arms hold at chest height.
Standing on 2–1–2 legs	ST	Two–legged stance–One-legged stance–Two-legged stance
Knee bend	KB	Two-legged stance–Bending knees–Two-legged stance.

measurements was finally performed so that the cycle of each trial consisted of 201 equidistant measurements (#1–#201).

The coordinates of external markers at legs and pelvis as well as the ground reaction forces were recorded in a fixed 'laboratory coordinate system' (Fig. 2). The marker positions relative to palpable bony landmarks were measured on the patients. The locations of joint centres and additional reference points relative to these landmarks, used for calculating the rotations, were determined using individual CT data. This allowed the calculation of the coordinates of joint centres and reference points relative to the laboratory coordinate system from the measured marker positions. These coordinates determine the positions and orientations of the body segments pelvis, thighs, shanks and feet in space.

Table 3
Numbers of averaged trials and patients[a]

Patient	HSR	PFL	KWR	IBL	NPA
Slow walking	1	5	5	—	HSR + PFL + KWR
Normal walking	8	5	8	5	HSR + PFL + KWR + IBL
Fast walking	5	4	5	—	HSR + PFL + KWR
Up stairs	6	2	6	(6)	HSR + PFL + KWR
Down stairs	4	1	6	—	HSR + PFL + KWR
Standing up	4	4	4	4	HSR + PFL + KWR + IBL
Sitting down	4	4	4	(4)	HSR + PFL + KWR
Standing on 2–1–2 legs	4	4	(4)	—	HSR + PFL
Knee bend	4	4	4	—	HSR + PFL + KWR

[a] The 'typical' patient NPA was calculated from the averages of 2 to 4 individual trials. Trials in parentheses were not used for NPA.

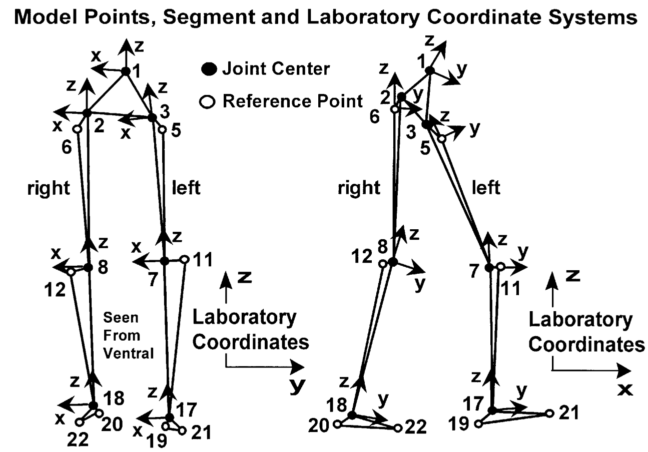


Fig. 2. Joint centres, reference points and coordinate systems for gait analysis. From the measured external marker locations (not shown) the coordinates of joint centres and additional reference points are calculated in the laboratory coordinate system. These points define segment coordinate systems, fixed to the distal skeletal segments of the joints. They describe the segment orientations in space. Inter-segmental forces and moments are caused by ground reaction forces, segment masses and their accelerations. They are calculated in the segment systems.

Table 4
Measured and calculated variables (see Fig. 2 and compact disc 'HIP98' for details)

Variable number	Variable	Body side	Coordinate system
1	Time	—	—
2–5	Contact foot—ground	Left + Right	—
6–15	EMG left and right leg	Left + Right	—
16–29	Magnitudes, positions + moments of ground reaction forces	Left + Right	Laboratory
30–32	Hip contact force F (negative components)	Left	Laboratory
33–35	Ground reaction force GL	Left	Thigh
36–38	Hip contact force F (negative components)	Left	Thigh
39–50	Inter-segmental force + moment from shank to foot	Left + Right	Shank
51–62	Inter-segmental force + moment from thigh to shank	Left + Right	Thigh
63–74	Inter-segmental force + moment from pelvis to thigh	Left + Right	Thigh
75–80	Inter-segmental force + moment from L5 to pelvis	Middle	Pelvis
81	Force at intervertebral disk L5-S1	Middle	Pelvis
82–90	Positions of points 1, 2, 3 at pelvis	Middle	Laboratory
91–102	Positions of points 5, 6, 7, 8 at thigh	Left + Right	Laboratory
103–108	Positions of points 11, 12 at shank	Left + Right	Laboratory
109–126	Positions of points 17, 18 at foot	Left + Right	Laboratory
127–132	Angles between pelvis and thigh	Left + Right	Anatomical
133–138	Angles between thigh and shank	Left + Right	Anatomical
139–144	Angles between shank and foot	Left + Right	Anatomical
145–146	Pelvis tilt sidewards and forwards	Middle	Anatomical
147–155	Transformation matrix of pelvis system	Middle	Laboratory
156–173	Transformation matrix of thigh system	Left + Right	Laboratory
174–191	Transformation matrix of shank system	Left + Right	Laboratory
192–209	Transformation matrix of foot system	Left + Right	Laboratory

The shapes of leg segments were described with simple geometric bodies, their sizes were measured on the individual patients (Deuretzbacher and Rehder, 1995). From these data and the average segment densities (Dempster, 1955) the segment masses and their centres of mass were calculated. Using these properties, the measured segment positions in space and their calculated accelerations plus the ground reaction forces, the inter-segmental forces and moments were computed. These loads have to be counter-balanced by joint contact forces and muscles. The inter-segmental loads (see CD) are reported in 'segment coordinate systems' (Fig. 2) which are connected to the proximal ends of the distal bones of the joints. At the hip joint the segment coordinate system is the same as the femur coordinate system (Fig. 1). The directions of the segment and laboratory axes x and y are different, even for a neutral standing position.

The angles between the segments are reported in the segment coordinate systems and additionally as anatomical angles. The given orientations of all segments relative to the laboratory coordinate system allow the hip contact force and all other inter-segmental loads to be transformed first to the laboratory coordinate system and from there to any other coordinate segment system (see CD). This way it is possible, for example, to determine the hip contact force directions relative to the pelvis.

2.5. Averaging of trials and patients

Two of the patients had instrumented implants at the right hip joint. All their data were mirrored to the left joint. For some activities the gait data of the leg without instrumented implant could not be evaluated. Before averaging gait data from the individuals their movement directions were standardised. For walking and stair climbing the laboratory coordinate systems were turned so that heel contact and toe off were placed on the axis x of the laboratory coordinate system (Fig. 2). For the other activities the line joining the hip joint centres at the beginning of the exercise was set parallel to the axis y.

A new method for averaging curves of varying shape was developed (details on CD). All cycle lengths are normalized and the curves are smoothed using Fourier series with n harmonics, starting with $n = 20$. If the number and sequence of remaining relative maxima and minima differs between the cycles, n is reduced and the procedure is repeated. The belonging extreme values from all curves are then shifted to average times within the common cycle and all curves are finally superimposed and averaged arithmetically. This method is aimed at calculating representative curves rather than averaging peak values which can slightly deviate from their exact arithmetic mean.

This averaging method was used to calculate average hip contact forces and gait data from several trials of the same patient, delivering an 'individual' average. Averages from several patients were then taken and averaged again, resulting in the 'typical' average for the fictional subject NPA. Due to lacking data the patients included in the activities of NPA were not always the same (Table 3) and this restricts the comparison between different activities.

2.6. Presentation of data

All data are available in detail from the CD. Two examples displaying several variables are shown in Fig. 3. Due to space restrictions this paper concentrates on the presentation of

1. Time courses of resultant contact force F and resultant implant moment M plus their components for the typical patient NPA. The component M_t is decisive for the torsion acting around the stem of the implant whereas the other components are of minor importance.
2. Peak contact forces F_p and peak torsional implant moments M_{tp} from single trials, individual and typical patients. F_p and M_{tp} are the highest values of F or M_t within an activity cycle (Fig. 4).

3. Results

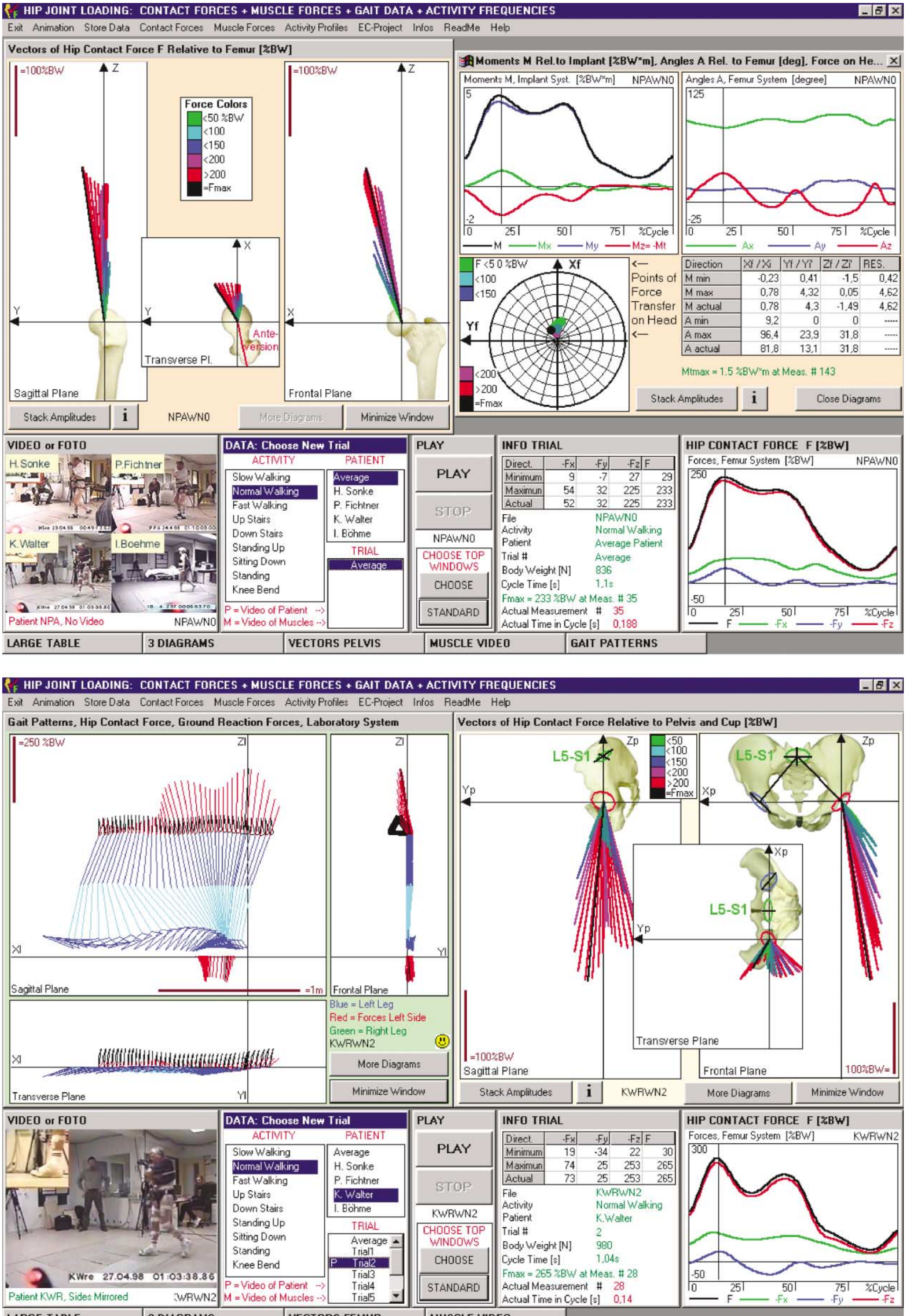
3.1. Averaging

Fig. 4 (top) gives an example of how the measurements were averaged. The contact force F from eight trials of patient KWR during normal walking and the individual average of this patient are displayed. The peak value of the average curve is $F_p = 242\%$ BW. Fig. 4 (bottom) repeats the average force curve of KWR and adds the components of F in the femur coordinate system. The individual averages from all four subjects were used to calculate the typical average of the fictional patient NPA (Fig. 5) with $F_p = 238\%$ BW.

3.2. Hip contact force

The intra-individual variations of contact force and gait patterns were mostly small for the cyclic activities of level and staircase walking (Fig. 4). The inter-individual differences, however, were often much larger (Fig. 5). During walking the expected double peak curves of F were observed in only two of the patients.

The contact forces F of the typical patient NPA and their components are charted in Fig. 6 for the nine investigated activities. The peak values F_p of the individual and average patients are listed in Table 5


HIP JOINT LOADING: CONTACT FORCES + MUSCLE FORCES + GAIT DATA + ACTIVITY FREQUENCIES
Exit Animation Store Data Contact Forces Muscle Forces Activity Profiles EC-Project Infos ReadMe Help
Vectors of Hip Contact Force F Relative to Femur [%BW]
=100%BW
Force Colors
<50 %BW
<100
<150
<200
>200
=Fmax
Ante-version
Transverse Pl.
Sagittal Plane
Frontal Plane
Stack Amplitudes
NPAWN0
More Diagrams
Minimize Window
Moments M Rel.to Implant [%BW*m], Angles A Rel. to Femur [deg], Force on He...
Moments M, Implant Syst. [%BW*m] NPAWN0
Angles A, Femur System [degree] NPAWN0
%Cycle
M Mx My Mz= -Mt
Ax Ay Az
Points of Force Transfer on Head
Direction Xf / Xi Yf / Yi Zf / Zi RES.
M min -0,23 0,41 -1,5 0,42
M max 0,78 4,32 0,05 4,62
M actual 0,78 4,3 -1,49 4,62
A min 9,2 0 0 -----
A max 96,4 23,9 31,8 -----
A actual 81,8 13,1 31,8 -----
Mtmax = 1,5 %BW*m at Meas. # 143
Stack Amplitudes
Close Diagrams
VIDEO or FOTO
H.Sonke P.Fichtner K.Walter I.Boehme
Patient NPA, No Video NPAWN0
DATA: Choose New Trial
ACTIVITY PATIENT
Slow Walking
Normal Walking
Fast Walking
Up Stairs
Down Stairs
Standing Up
Sitting Down
Standing
Knee Bend
Average
H. Sonke
P. Fichtner
K. Walter
I. Böhme
TRIAL
Average
P = Video of Patient -->
M = Video of Muscles -->
PLAY
STOP
NPAWN0
CHOOSE TOP WINDOWS
CHOOSE
STANDARD
INFO TRIAL
Direct. -Fx -Fy -Fz F
Minimum 9 -7 27 29
Maximum 54 32 225 233
Actual 52 32 225 233
File NPAWN0
Activity Normal Walking
Patient Average Patient
Trial # Average
Body Weight [N] 836
Cycle Time [s] 1,1s
Fmax = 233 %BW at Meas. # 35
Actual Measurement # 35
Actual Time in Cycle [s] 0,188
HIP CONTACT FORCE F [%BW]
Forces, Femur System [%BW] NPAWN0
F -Fx -Fy -Fz
LARGE TABLE
3 DIAGRAMS
VECTORS PELVIS
MUSCLE VIDEO
GAIT PATTERNS
HIP JOINT LOADING: CONTACT FORCES + MUSCLE FORCES + GAIT DATA + ACTIVITY FREQUENCIES
Exit Animation Store Data Contact Forces Muscle Forces Activity Profiles EC-Project Infos ReadMe Help
Gait Patterns, Hip Contact Force, Ground Reaction Forces, Laboratory System
=250 %BW
Sagittal Plane
=1m
Frontal Plane
Transverse Plane
Blue = Left Leg
Red = Forces Left Side
Green = Right Leg
KWRWN2
More Diagrams
Minimize Window
Vectors of Hip Contact Force Relative to Pelvis and Cup [%BW]
L5-S1
=100%BW
Sagittal Plane
Transverse Plane
Frontal Plane
100%BW=
Stack Amplitudes
KWRWN2
More Diagrams
Minimize Window
VIDEO or FOTO
KWre 27.04.98 01:03:38.86
Patient KWR, Sides Mirrored
DATA: Choose New Trial
Trial1
Trial2
Trial3
Trial4
Trial5
INFO TRIAL
Direct. -Fx -Fy -Fz F
Minimum 19 -34 22 30
Maximum 74 25 253 265
Actual 73 25 253 265
File KWRWN2
Activity Normal Walking
Patient K.Walter
Trial # 2
Body Weight [N] 980
Cycle Time [s] 1,04s
Fmax = 265 %BW at Meas. # 28
Actual Measurement # 28
Actual Time in Cycle [s] 0,14
HIP CONTACT FORCE F [%BW]
Forces, Femur System [%BW] KWRWN2
VECTORS FEMUR

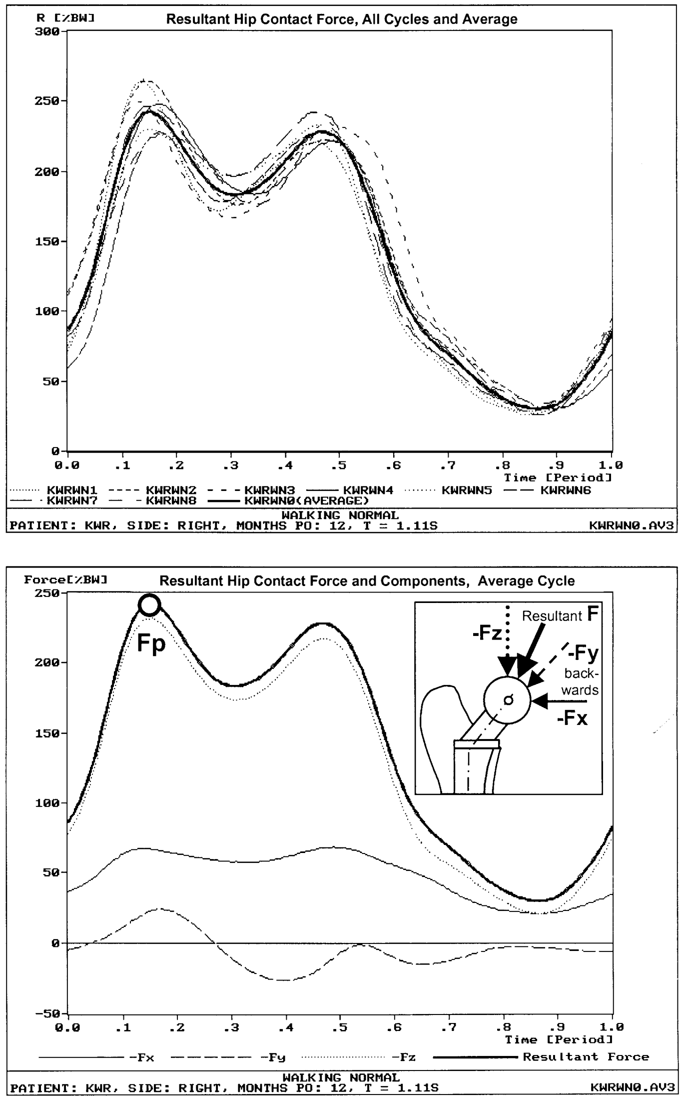


Fig. 4. Contact force F of single trials and individual patient during normal walking. *Top*: Hip contact force F in % BW from eight trials of patient KWR (thin lines) and the individual average of this patient (thick line). *Bottom*: Individual average of F from top diagram and its components $-F_x$, $-F_y$, $-F_z$. The highest value is the peak force F_p.

which additionally contains the average cycle times T. F_p doesn't differ much between slow, normal and fast walking. This may be misleading, however, because the patient groups were different (Table 3) and the speed of slow and normal walking was nearly the same (Table 2). The curves of NPA during the stance phases of walking and going upstairs look very similar at first glance. The component $-F_y$, however, which causes much of the implant torque, is larger when going upstairs. Downstairs the peak force slightly exceeds that from going upstairs. Standing up from a chair loads the hip joint more than sitting down but much less than walking. The rotating component $-F_y$ is very small when sitting down. Standing on one leg let the contact force rise to

Fig. 3. Data from hip contact force measurement and gait analysis. (See next page) Screen dumps from compact disc 'HIP98'. Variable notations see Fig. 1. **Upper diagrams**: Typical patient NPA during one cycle of normal walking. *Top left*: Vectors of hip contact force F relative to femur, *Top right*: Implant moment M, contact force angles A, points of force transfer in cup, table. Bottom from left: Photographs of averaged patients at instant of maximum contact force. Selected activity, patient and trial. Animation window. Table with contact force F and information about trial. Diagram of F and its components; the display time can be chosen by moving the cursor line. **Lower diagrams**: Patient KWR during one trial of walking down stairs. *Top left:* Stick diagrams from gait analysis. *Top right:* Contact force vectors F relative to pelvis. *Bottom left*: Synchronised video of patient.

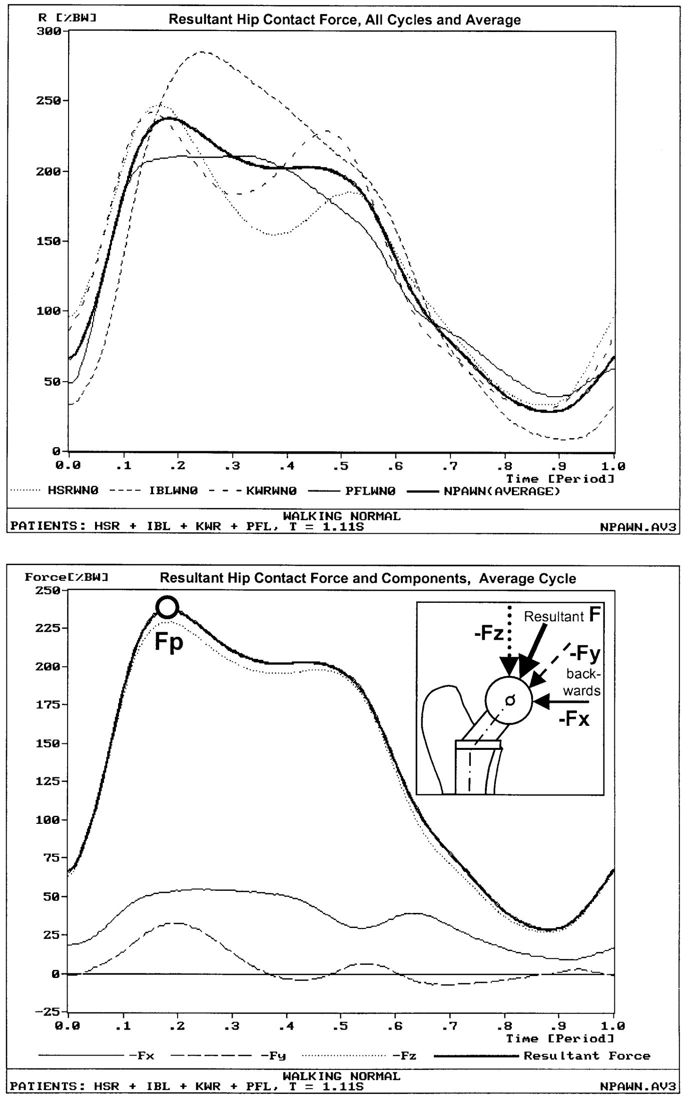


Fig. 5. Contact force F of individual patients and typical patient NPA during normal walking. *Top*: Individual averages of hip contact force F in % BW from four patients (thin lines) and the typical average of patient NPA (thick line). *Bottom*: Typical average of F from top diagram and its components $-F_x, -F_{y'}, -F_z$.

nearly the same peak value as walking. Knee bends are not very strenuous for the hip joint.

Fig. 7 charts the peak contact forces F_p from the typical patient NPA, the highest average of individual patients and the two extreme single trials. From the difference between the columns for 'Average Patient' and 'Max. Patient' one can estimate the variations between the patients. The different heights of white and black columns represent the spread between all investigated single trials.

3.3. Contact force directions

Fig. 8 assembles the vectors of the contact force F from the typical patient NPA as seen in the frontal and transverse planes of the femur. The force directions in the frontal plane (top diagrams) are very similar during all activities and their variation is remarkably small. Small forces act more from medial than large ones. The indicated angle A_y of the peak force F_p is in the extremely small range of 12–16° for all activities except standing on one leg when it is 7°.

The angle A_z in the transverse plane (Fig. 8, bottom diagrams) varies more than A_y. During activities which cause high forces, i.e. for standing, level and staircase walking, A_z increases with the magnitude of F. The indicated directions A_z of the peak force F_p are in the range of 28–35° when standing, walking and going downstairs. For walking upstairs $A_z = 46°$ is larger.

3.4. Implant moments

Similar observations as for the contact force F (Fig. 6) are made when the implant moments M are analysed

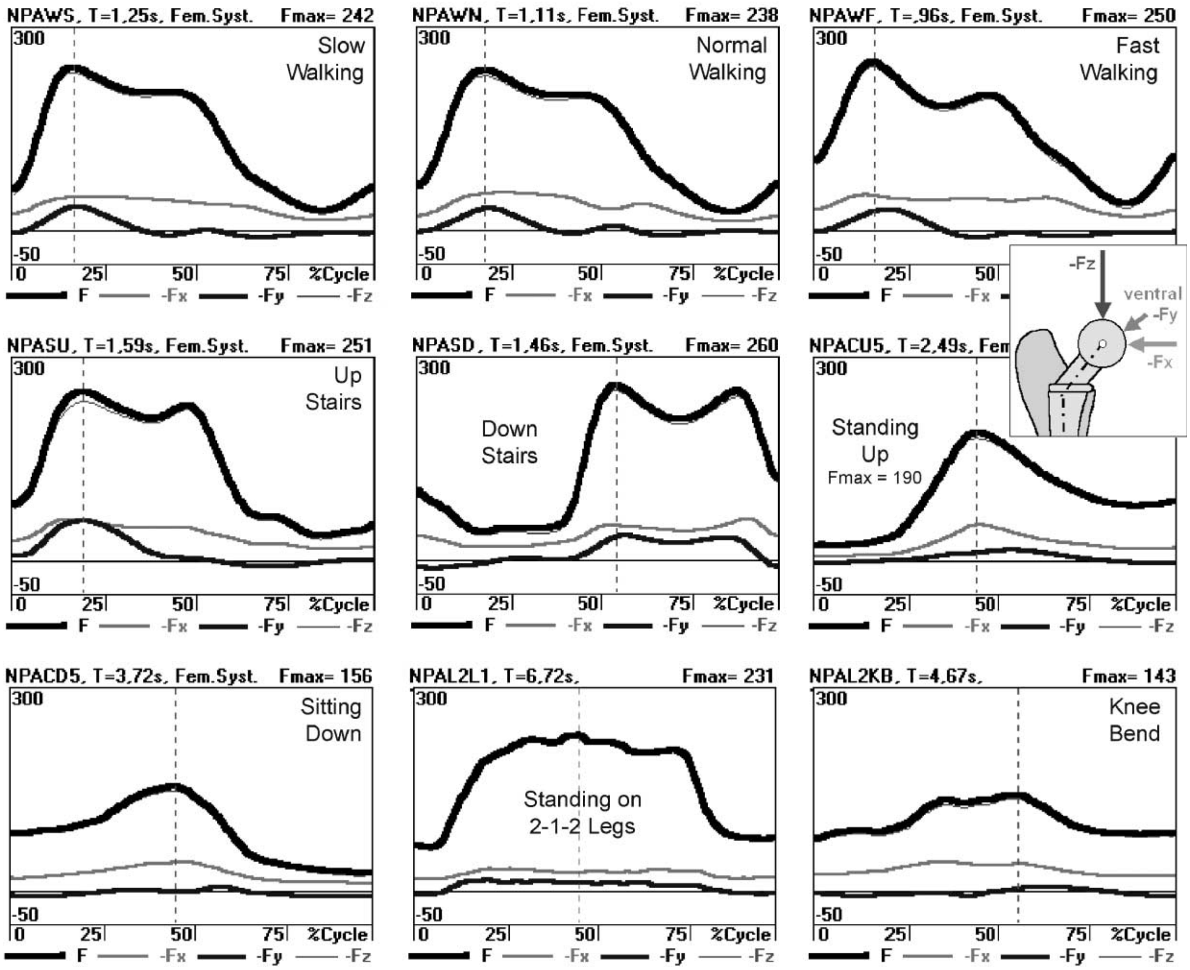


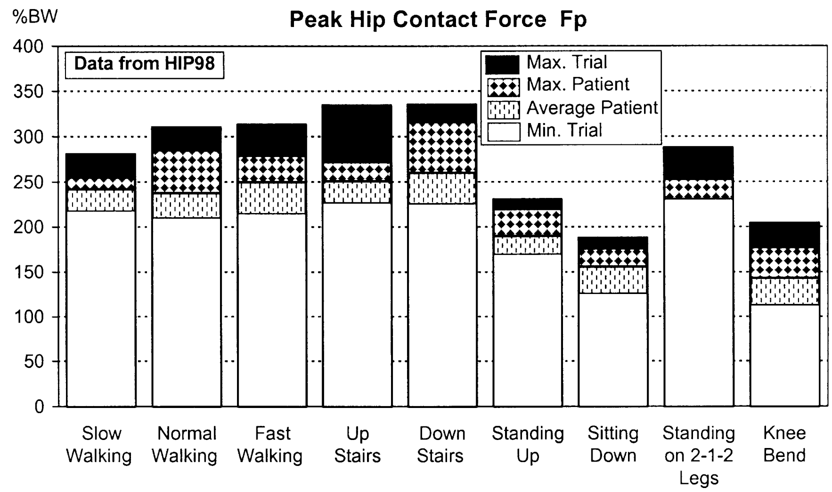
Fig. 6. Contact force F of typical patient NPA during nine activities.Contact force F and its components $-F_x, -F_y, -F_z$. F and $-F_z$ are nearly identical. The scale range is -50–300% BW. Cycle duration and peak force $F_p = F_{max}$ are indicated in diagrams.

Table 5
Peak loads of single and average patients, cycle times and body weight of average patient[a]

Patient and activity	Peak hip contact force F_p (% BW)					Peak torsional implant moment M_{tp} (% BW m)					Cycle time (s)	Body weight (N)
Patient	HSR	PFL	KWR	IBL	NPA	HSR	PFL	KWR	IBL	NPA	NPA	NPA
Slow walking	239	255	244	—	242	1.72	1.65	1.71	—	1.64	1.25	847
Normal walking	248	211	242	285	238	1.82	1.25	1.64	1.55	1.52	1.11	836
Fast walking	279	218	275	—	250	1.91	1.21	1.94	—	1.54	0.96	847
Up stairs	265	227	272	(314)	251	2.25	1.82	2.96	(2.92)	2.24	1.59	847
Down stairs	263	226	316	—	260	1.83	1.63	2.33	—	1.74	1.46	847
Standing up	181	208	182	220	190	1.18	0.77	1.03	1.01	0.88	2.49	836
Sitting down	176	153	149	(199)	156	0.91	0.37	0.65	(0.75)	0.47	3.72	847
Standing on 2–1–2 legs	253	223	(369)	—	231	1.64	0.96	(1.55)	—	1.17	6.72	920
Knee bend	177	117	147	—	143	0.67	0.58	0.83	—	0.51	6.67	847

[a] Peak hip contact forces F_p in % BW and peak torsional implant moments M_{tp} in % BW m. HSR, PFL, KWR, IBL are 'individual' averages from several trials of the patients, NPA is the 'typical' average from 2 to 4 patients (Table 3). Data in parentheses were not used for calculating NPA. The body weight of NPA depends on the number of averaged patients (Table 3).

(Fig. 9). The curves of F and M have very similar shapes for the same activity which reflects the limited variation of the force directions. Whereas the peak torsional implant moment $M_{tp} = -M_{z'p}$ is between 1.72 and 1.91% BW m for the typical patient for walking, it is 2.25% BW m when going upstairs (Table 5). Furthermore the variation between patients and trials is much larger while going upstairs than during walking (Fig. 7).



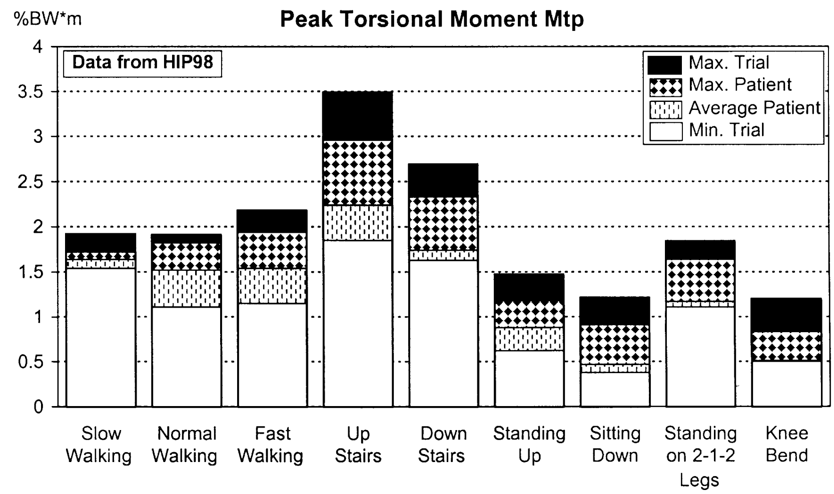


Fig. 7. Average, minimum and maximum peak values of contact force F_p and torsional moment M_{tp}. Peak values from nine activities. Max. Trial = Highest value from all investigated trials. Max. Patient = Highest individual average from all patients. Average Patient = Value of typical patient NPA. Min. Trial = Lowest value from all investigated trials. *Top*: Contact force F_p in % BW. *Bottom*: Torsional implant moment $M_{tp} = -M_{z'p}$ in % BW m.

Implant torsion is small when sitting down, standing up or during knee bend. When standing on one leg it is comparable to walking.

4. Discussion

The presented data base of hip joint loading is the most complete today. Nevertheless, the number of patients is too small to use statistical methods for further data evaluation. The inter-individual variation between patients (Fig. 5) indicates that the average loads would change if more or other patients had been included in this study. For getting as representative data for each activity as possible, all patients were included in the average patient NPA (Table 3) instead of only HSR and PFL, who were measured during all activities. The varying patient group has to be considered, however, when comparing the different activities of NPA. Slightly higher peak forces during slow as compared to normal walking (Table 5), for example, are caused by the exclusion of patient IBL rather than by the walking speed.

Contact forces with a double peak curve during walking (Fig. 4), similar to the ground reaction force, are usually regarded as 'normal'. In two patients single peak contact forces have been observed, however (Fig. 5), while the belonging ground reaction force showed the usual double peak pattern (see CD). Obviously there is no strict relation between both. We could not relate the occurrence of such single peak forces during walking to the anteversion angle (Table 1), to the trial-to-trial variability of contact forces, to the side at which the prosthesis was implanted or to any other factor. At least in patients with hip implants the inter-individual variation of contact loads seems to be larger than expected previously. Because no criterion exists for 'normal' contact force patterns and because

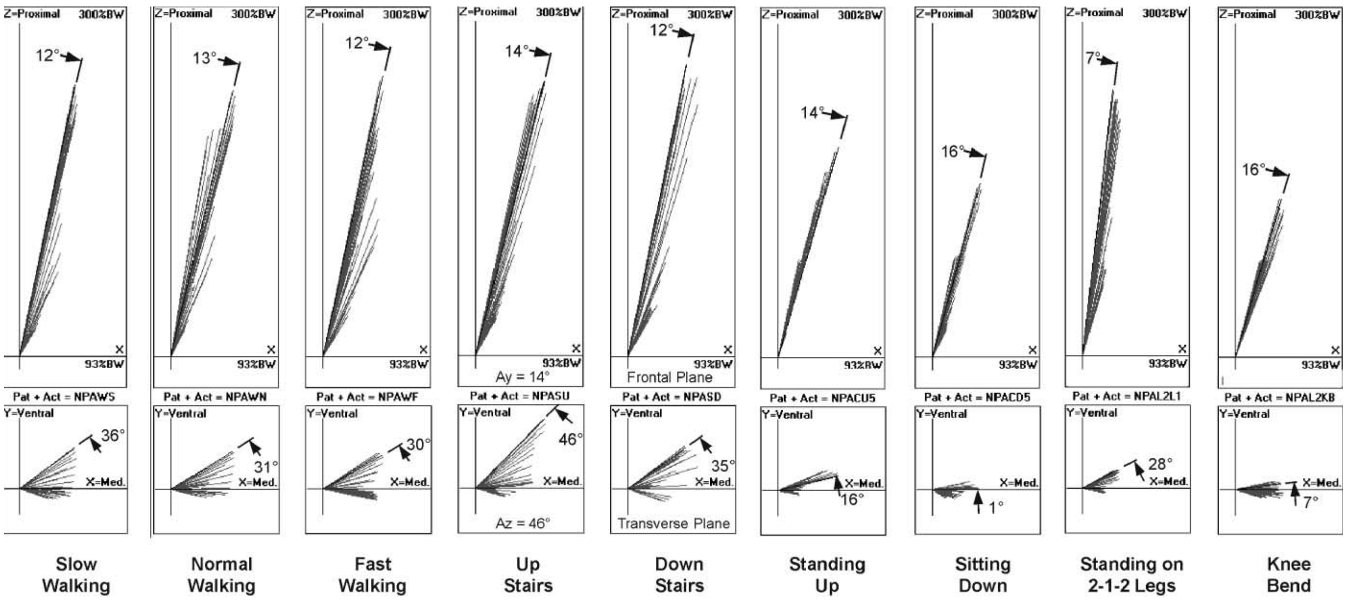


Fig. 8. Contact force vector F of typical patient NPA during nine activities. The z-scales go up to 300% BW. *Upper diagrams*: Force vector F and direction A_y of F in the frontal plane. *Lower diagrams*: Force vector F and direction A_z of F in the transverse plane.

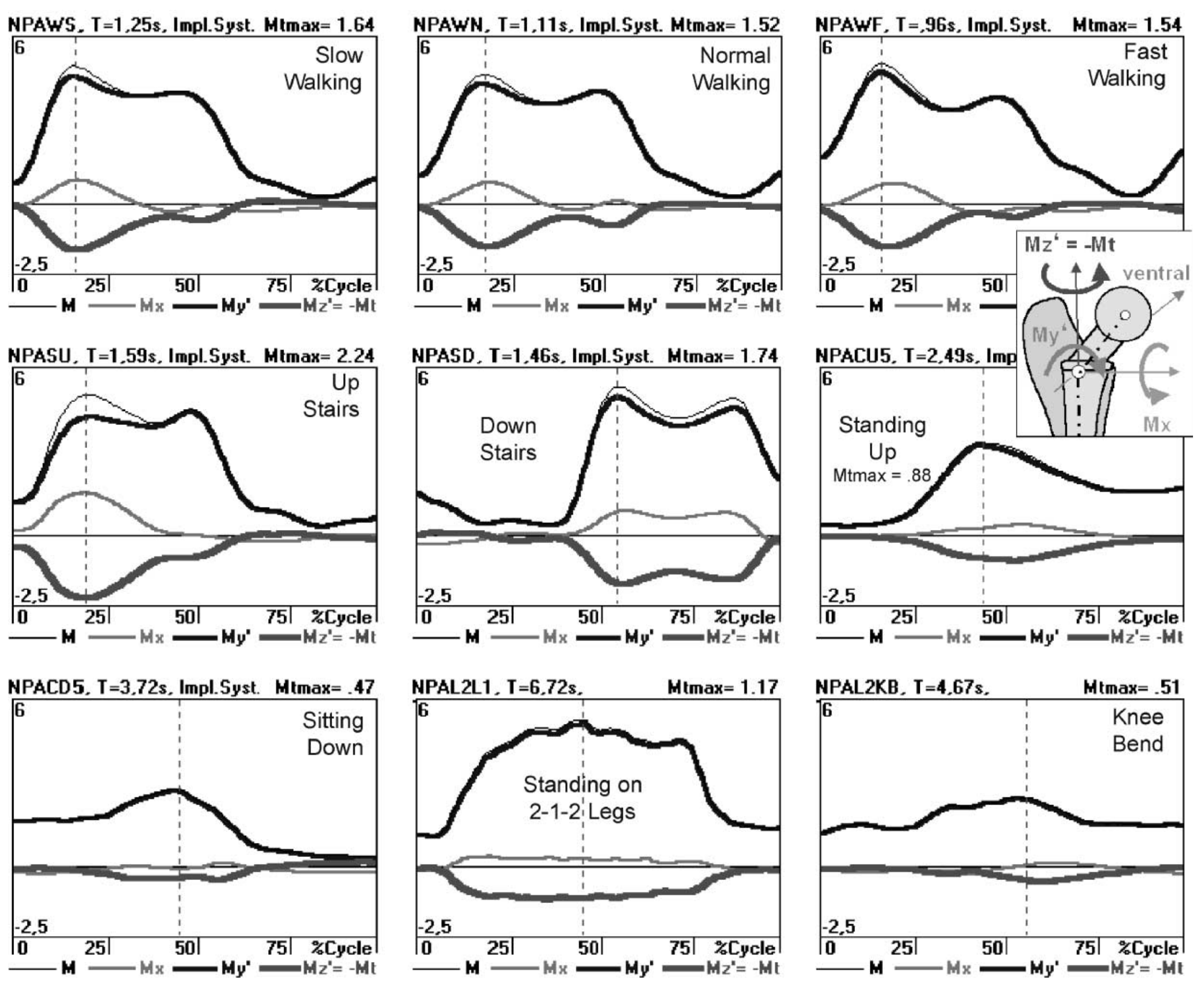


Fig. 9. Implant moment M of typical patient NPA during nine activities. Implant moment M and its components $M_x, M_{y'}, M_{z'} = -M_t$ in % BW m. M and $M_{z'}$ are nearly identical. The scale range is -2.5–6% BW m. Negative M_z cause inwards torsion around the implant stem. Cycle duration T and peak torsional moment $M_{tp} = M_{tmax} = -M_{zp}$ are indicated in diagrams.

the average data are mainly aimed at testing hip implants in a realistic way, it seems justified to include all patients in the average patient NPA.

The average peak forces F_p of the patients during normal walking at about 4 km/h were between 211 and 285% BW (Fig. 5, Table 5). This range is similar to those measured by other authors with instrumented implants (Rydell, 1966a, b; English, 1977, 1978; English and Kilvington (1979); Davy et al., 1988, 1990; Kotzar et al., 1988). Higher forces of 307 and 324% BW at 3 km/h have previously been found in the two joints of our first patient EB with instrumented implants (Bergmann et al., 1993). This shows that the variations between patients are considerable as can also be seen from the deviating curve shapes in Fig. 5. Even larger differences between the patients were observed for non-cyclic activities like standing on one leg or standing up from a chair (see CD). Probably walking, which is the most frequent activity with high contact forces, is best 'optimised' anatomically and by training with regard to low loads levels and/or energy consumption. If this assumption is true, it can be expected that load or energy criteria, used for the calculation of internal forces, will deliver best results for walking but less realistic data for other activities.

The peak contact force in a patient with disturbed gait patterns has previously been found to be as much as 409% BW during walking (Bergmann et al., 1993, patient JB), which is far greater than the limits now observed. This supports the opinion that dysfunction of one muscle increases the joint contact force, because a part of the required joint moment is taken over by other muscles with unfavourably short lever arms and therefore higher forces.

The contact force directions in the frontal plane vary only slightly during the activities reported here and are nearly the same when the force reaches its peak value within a loading cycle (Fig. 8). Unpublished own data from more patients and activities prove that this is generally the case. This supports the hypothesis that anatomy and function of the mechanical system consisting of femur, pelvis and acting muscles is optimised in a way that limits the highest bending stresses in the femoral shaft (Pauwels, 1965; Duda et al., 1997). High forces in the transverse plane act more from anterior than lower ones (Fig. 8). A similar biological optimisation strategy may exist with regard to minimised torsion of the natural femur or the implant. The anteversion of an implant therefore influences its fixation stability.

Due to the relatively poor torsional stability of the stem fixation, torque may endanger the implant stability more than bending. When going upstairs the average torque is 23% higher than during normal walking (Table 5), but when walking very fast the torque will be probably in the same range (Bergmann et al., 1995b). It must be noted, however, that the variation between patients and trials is very large when going upstairs (Fig. 7). The absolutely highest observed implant torque when going upstairs was 83% larger than during normal walking. Obviously climbing stairs, which typically occurs not as frequently as walking, is performed in a less optimised way than walking. It can therefore not be excluded that going upstairs endangers the implants of patients with unsafe walking ability if the fixation stability of their implants is poor. A stem design with good torque resistance is therefore a fundamental requirement.

The low forces and moments when using a chair or during knee bends show that such activities are not decisive when testing the fatigue strength of the implant fixation. From the hip joint loads measured in seven patients during many other common activities of daily living no implant forces or moments more critical than during fast walking or stair climbing, respectively, have been detected (unpublished). One exception was stumbling, when extremely high forces of up to 870% BW acted (Bergmann et al., 1993). The most realistic way to test hip implants with regard to their fixation stability will therefore be to apply large numbers of loading cycles as during walking, a smaller number of cycles as during going upstairs and occasional forces as during stumbling. Propositions for test scenarios of different complexity will be made in a separate paper.

In this paper data presentation and discussion have been concentrated on the contact forces in the hip joint. The CD accompanying this paper contains much more data from these investigations. The documented movements in the hip joint allow one to calculate the force directions relative to the acetabulum. This opens the prospect for testing the stability of acetabular cups and other aspects of the musculo-skeletal loading of the pelvis. The data can also serve for setting up more realistic friction and wear tests than those used today. The included data about the activities of hip joint muscles (Heller et al., 2001) and the frequencies of everyday activities (Morlock et al., 2001) will serve as basis to define realistic test scenarios for hip implants and investigate the role of normal and disturbed muscle functions for the loads acting in the hip region.

Acknowledgements

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