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HIP STRESS DURING LIFTING WITH BENT AND STRAIGHT KNEES

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ABSTRACT. "Correct" body mechanics during lifting are believed to protect the back by employing knee and hip flexion while keeping the back straight. Lower limb joint stress, however, has been largely ignored. We compared hip cartilage contact stress during "leg lifting" with "back lifting" (lifting with bent or straight knees, respectively) in a subject fitted with a pressure instrumented hip endoprosthesis. Synchronized pressure data and whole-body kinematics and kinetics were collected simultaneously while the subject lifted an 11.8 kg mass from the floor to waist level. The highest pressure, 13.7 MPa, occurred during leg lifting at the antero-lateral femoral head transducers opposed at maximum hip flexion by the postero-superior quadrant of acetabular cartilage. In back lifting, the highest pressure, 11.5 MPa occurred in the supero-lateral aspect of the head, which during hip extension was opposed by the posterior quadrant of the acetabulum. Maximum pressures and hip torques occurred simultaneously with peak hip flexion, during the initial lifting of the burden from the floor. Acetabular contact pressures during leg lifting were on average twice as great as during back lifting, and both techniques generate much greater hip stress than gait (typically 4–6 MPa). Degenerative changes in the articular surface of the acetabulum occur primarily on the postero-superior aspect, corresponding to the locations of peak contact pressures in the present *in vivo* data. Thus leg lifting puts more stress on the postero-superior region, and probably contributes to more hip cartilage degeneration, than does back lifting. We conclude that although leg lifting may mechanically protect the back, it substantially increases hip cartilage stress.

Key words: biomechanics, hip stress, lifting technique.

Manual lifting is the most common cause of back pain, accounting for one third of all industrial injuries (28), causing back impairment ranging from minor back strain to vertebral disc rupture (2, 23, 38). To determine the musculoskeletal factors responsible, biomechanical studies of lifting have focused on spinal load and back muscle activity (15, 16, 18, 26), intradiscal pressure (29) or the torques or moments imposed on the lower back (28). Bejjani et al. (6), reported decreased lower back loads during lifting with the knees bent and back straight (leg lifting), compared to lifting with the knees straight (back lifting) which placed the burden further from the trunk. Bendix & Eid (5) found that back lifting presents a greater risk of back injury than leg lifting. Consequently, ergonomicians now widely advocate leg lifting to prevent and reduce back pain among workers and patients experiencing low back pain. "Lift with your legs and not with your back," is the common suggestion, implying that decreased load on the back increases hip and knee loads (27, 28).

Little data on hip loading during leg lifting exist to support this common wisdom. Indeed, hip muscle and joint force estimates based on external kinematic/kinetic data, have produced widely varying results during gait and other ostensibly identical tasks (11, 12). Measurements from instrumented hip prostheses, beginning with Rydell in 1966, provide the most accurate quantitative *in vivo* hip force (4, 14, 33), and acetabular contact pressures (19, 21, 22, 25, 36) data. To date, however, none have reported *in vivo* hip stresses during lifting.

Static estimates of lifting kinetics (31) suggest there are negligible hip compression force differences between back and leg lifting. The magnitude of femoral head force in back lifting was reported to be greater (2.7% Body Weight [BW] vs 2.6% BW in back and leg lifting, respectively), due to the effective muscle moment arms around the joint (28, 31). This simplified model did not consider three-dimensional kinematics: ab/adduction and transverse rotations were ignored, as were the muscle co-contraction forces that stabilize the joint. Toussaint et al. (37), report high EMG activity of the biarticular semitendinosus and biceps femoris muscles during leg lifting. In contrast, Hagen et al. (20) found high activity of vastus lateralis and low activity of biceps femoris in leg lifting. Nemeth et al. (30) predicted the hip compression force was influenced strongly by the degree of hip flexion, due to both moment arm lengths and muscle load sharing. Anderson & Winters (1), examining forward and backward bending activities, credited passive support structures for decreasing back muscle activity at full flexion (17). Oxygen consumption in leg lifting is reported to be higher than in back lifting (20), and patients tend to revert to back lifting when the load is heavy (28, 34). Moreover, the relatively high risk of hip arthritis in manual lifting workers (35, 38) may be related to muscle co-contraction about the hip joint (21, 22, 26, 37). In summary, hip stresses during leg and back lifting have not been measured *in vivo* despite their importance to ergonomics and rehabilitation.

The purpose of the present study was to test the hypothesis that leg lifting increases hip stress and back lifting decreases hip stress. The results from this study are also intended to advance understanding of the mechanics of lifting, and improve the design of rehabilitation programs for lifting and work activities of manual materials-handling workers.

MATERIAL AND METHODS

Subject

In December 1991, a right-handed white 82-year-old male, 59 kg in weight and 1.6 m in height, sustained a Garden III left femoral neck fracture which required femoral head and neck replacement using an endoprosthesis. After excluding hip disease, matching the endoprosthesis size with the patient's femoral head and obtaining informed consent, an Austin-Moore-like femoral hemiarthroplasty instrumented with pressure transducers was implanted using the standard posterolateral approach (Kocher incision) through the short external rotators without disturbing the greater trochanter or the abductors (21). The exposed acetabular cartilage was radiographically and visually ascertained to be normal. Methymethacrylate was used to secure the prosthesis to permit early mobilization.

Rehabilitation, while recording pressure data, commenced the first day after the operation according to the hospital's usual routine for hip endoprosthesis patients (36). After hospital discharge, repeated and synchronized measurements of the acetabular contact stress, and kinetics and kinematics during gait and other functional activities were obtained. Observation of these functional performance data revealed

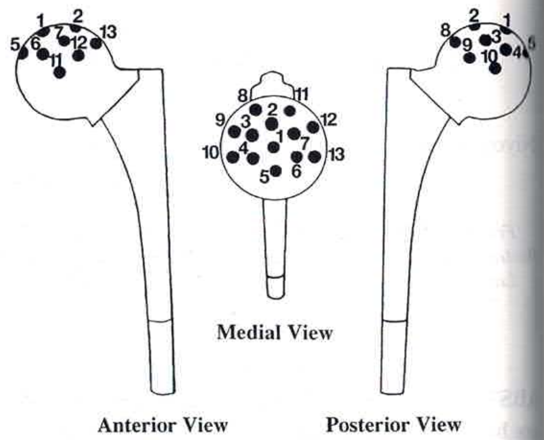

Fig. 1. Schematic rendering of the instrumented left hip endoprosthesis and its 13 pressure transducer locations.

that the patient had resumed near normal or normal locomotor function by 3 months post-operatively. The subject was free of pain, had no observable limp, walked 1 km and up and down three flights of stairs to his dwelling each day.

Instrumented prosthesis

The implanted prosthesis is devised from the same cobalt chromium, bio-compatible alloy used in the standard Austin-Moore device. The upper hemisphere contains 13 pressure transducers, arranged with one central transducer surrounded by two consecutive rings of six transducers (Fig. 1). The pressure-sensing diaphragms are integral with the acetabular-contacting hemisphere surface and deflect 0.00028 mm for each MPa of applied pressure. Diaphragm deflection is transferred by a pin and measured by the corresponding deflection of a single-silicon-crystal cantilever beam, on which is fused a semiconductor 4-arm strain bridge. Otherwise similar to the first pressure-instrumented endoprosthesis (9, 10, 21, 36), the present prosthesis adds a thermistor to measure hip temperature as it may affect the pressure signal. Inside, the hermetically-sealed hemisphere electronic circuits sequentially sample the strain gauge outputs and multiplex the frame of data (13 pressures and temperature) with a frame rate of 500 Hz as a pulse-amplitude modulated radio-telemetered signal in the 100 MHz FM band. A silver-wire pair carries the signal from the hemisphere to the antenna at the distal stem end, which also receives power from an external 100 KHz generator via an induction-coil garter worn around the thigh during acquisition. Excluding batteries from within the prosthesis eliminates their potential toxic products and extends power supply availability as long as desired.

Kinetic and kinematic data

Kinetic and kinematic data were aquired from 2 forceplates (Kistler 9281A, Switzerland) and four opto-electronic cameras (Selspot, Sweden), sampled simultaneously at 150 Hz. The forceplates measured the three orthogonal forces and three moments of foot-floor contact. The cameras sampled 980 nm wavelength infra-red signals from light

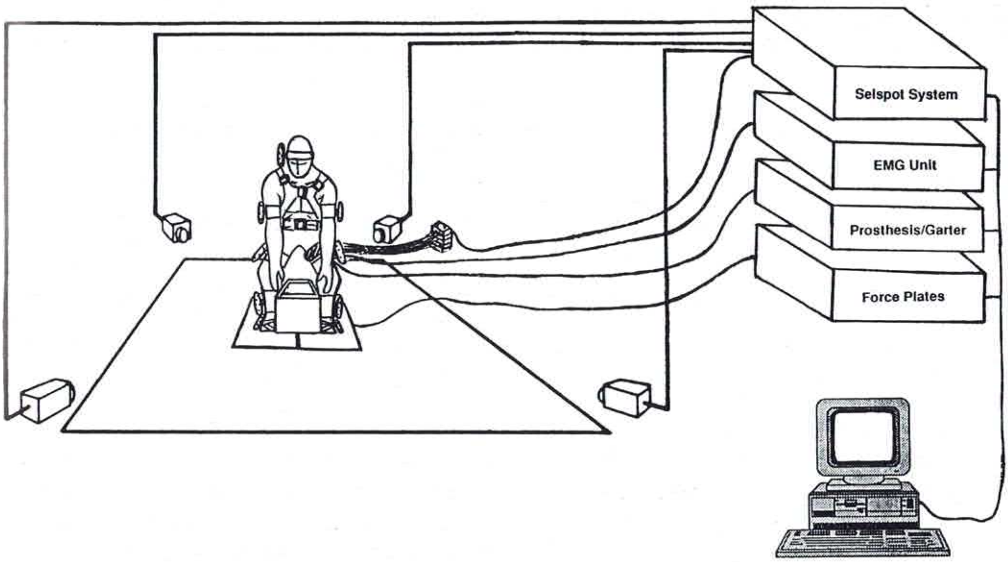

Fig. 2. Data collection hardware configuration.

emitting diodes (LEDs) mounted on plexi-glass arrays attached to the head, trunk, pelvis and extremities (Fig. 2). The three-dimensional orientations and positions of these 11 body segments are attained after processing by TRACK© (3) and Massachusetts General Hospital (MGH) software (3, 24, 32). The forceplate kinetics, and translation and rotation of each segment and their time derivatives together with segment inertial parameters (mass, center of gravity and moment of inertia) are used to estimate the forces and moments across intersegmental joints using NEWTON© software (3).

Endoprosthesis data

An external receiver unit demodulates the prosthesis FM signal, and assures synchronization with the other data. Data were acquired via three A/D channels of a 486 PC computer, sampling at 8 KHz. Channel 1 is prosthesis output, composed of 18 measurements: 13 pressures, temperature and parameters used to calibrate and synchronize the data, multiplexed onto one channel. Channel 2 acquires processing and alarm data which detect when pressure and temperature data are outside acceptable ranges. Channel 3 records redundant force plate data and TRACK© synchronization signals, to insure forces and subsequently estimated torques time-match the measured pressure data.

Procedure

The subject lifted a box with two handles from the floor to a bench 0.94 m high, the subject's umbilicus height. Four lifting techniques were studied; the subject put the box on the bench after each trial:

— Back lifting: the box was initially 2.5 cm in front of the toes
— Leg lifting close: the box was initially between the feet, i.e. close to the body
— Leg lifting far: the box was initially 2.5 cm in front of the toes, as in back lifting.
— Leg lifting close with step: the box was initially between the feet, i.e. close to the body; after lifting the box, the subject took a left step forward toward the bench.

Before commencing the lift, the subject stood with one foot on each forceplate, but without other restrictions on foot placement. Foot movement from the initial location was not allowed until completion of the task (except in leg lifting with step). The box weighed 0.5 kg and was 0.33 m wide, 0.33 m deep, 0.28 m high. Inside the box was a weight-lifter's disc (11.34 kg, 30 cm diameter disc). All lifts were performed at the subject's preferred speed, with 2–3 minutes rest between lifts. Instruction and practice were provided before data collection trials.

Data analysis

Hip contact pressure data were collected concurrently with kinematic and kinetic data; pressure data were plotted against the relevant hip, knee and back joint motions, torques and other data using MGH-developed software, while a time-synchronized graphic display of the subject was examined to insure data integrity. Maximum pressures during any trial, irrespective of lifting phase, were determined, then kinematic and kinetic correlates were obtained to determine the lifting phase and to interpret the pressure patterns. Contact pressure differences of 0.2 MPa or greater are considered significant and well beyond measurement error (10, 25).

RESULTS

The results are summarized in Table I and Figs 3–5. During lifting, only the four transducers on the superior quadrant changed in pressure magnitude more than 1 MPa. This pattern prevailed in both leg lifting

Table I. *Maximum pressures in MPa in experiment 1 (5 months post-implant), experiment 2 (16 months post-implant) and experiment 3 (28 months post-implant), from the four highest-reporting transducers in a given experiment*

	Experiment 1				Experiment 2				Experiment 3			
Transducer No.	2	3	6	7	1	2	3	7	1	2	3	7
Back lift	4.90	4.45	1.80	3.12	11.49	4.05	6.07	2.78	10.20	2.57	4.99	2.66
Leg lift/WF	5.43	3.67	2.34	8.16	7.36	5.46	8.66	5.23	3.26	2.94	6.60	6.39
Leg lift/WC	7.92	4.07	3.28	11.73	<1.0	10.02	9.91	13.71	<1.0	11.75	8.18	12.34
Leg lift/Step	7.23	4.58	2.86	9.95	na	na	na	na	5.15	8.38	8.24	13.08

See Fig. 1 for transducer locations. WF = weight far; WC = weight close; na = not available because test not performed in 16 week post-implant experiment

while the hip was in flexion, abduction and external rotation (to admit the box between the legs), and in back lifting while the hip was in flexion, adduction and slight internal rotation. Overall, the highest pressures occurred during leg lifting on transducer 7 (Table I; Fig. 3b, c and d). In back lifting, throughout the task, the highest pressures are reported by transducer 1 or 2 (Table I, Fig. 4). In leg lifting trials, the maximum pressure shifted to transducer 7 at the time of maximum hip flexion (Fig. 5), which coincided with the lifting forces imposed by the box as it left the floor. Maximum contact pressures were 19–139% greater in leg lifting than in back lifting (Table I).

The maximum hip contact pressure was higher with leg lifting close than with leg lifting far by 44–78% (Table I).

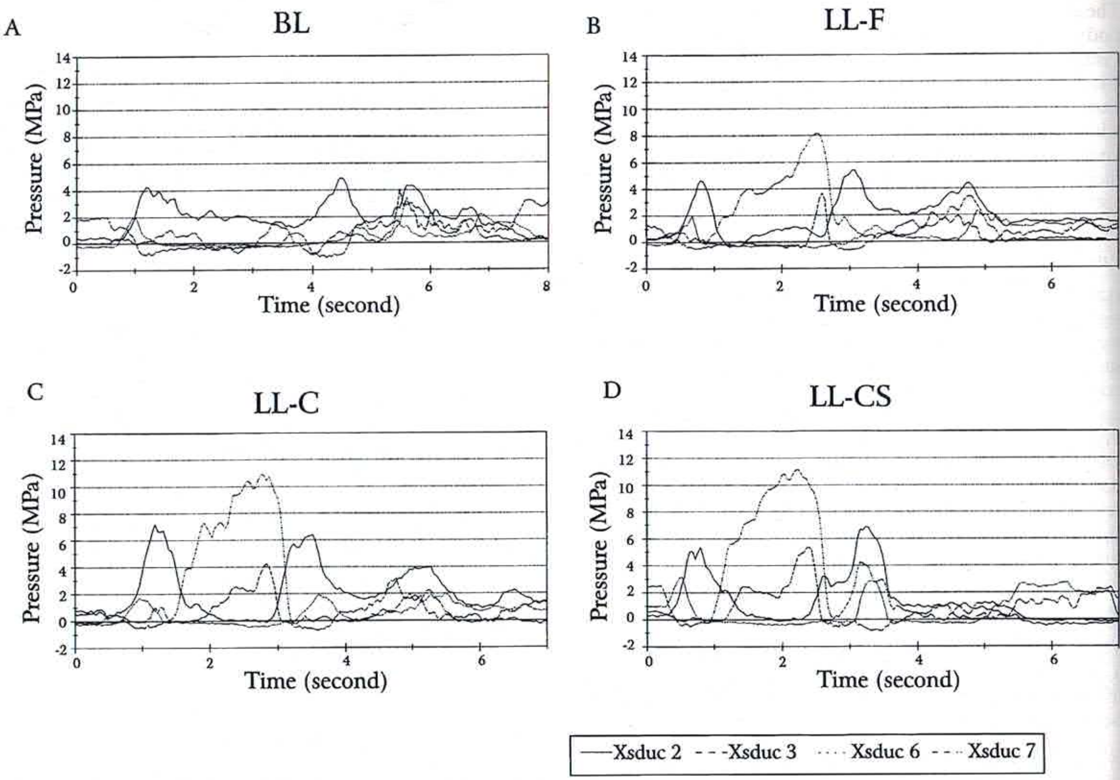

Fig. 3. Representative pressure data from the four transducers whose peak values exceed 1 MPa during lifting. (A) Back lifting; (B) leg lifting with the weight far (burden in front of feet); (C) leg lifting with weight close (burden between feet); (D) leg lifting with weight close, then stepping forward with the left foot. Note the first peak's similarity to 5C; second peak at about 3.2 seconds occurs as weight is born on the left limb following the step.

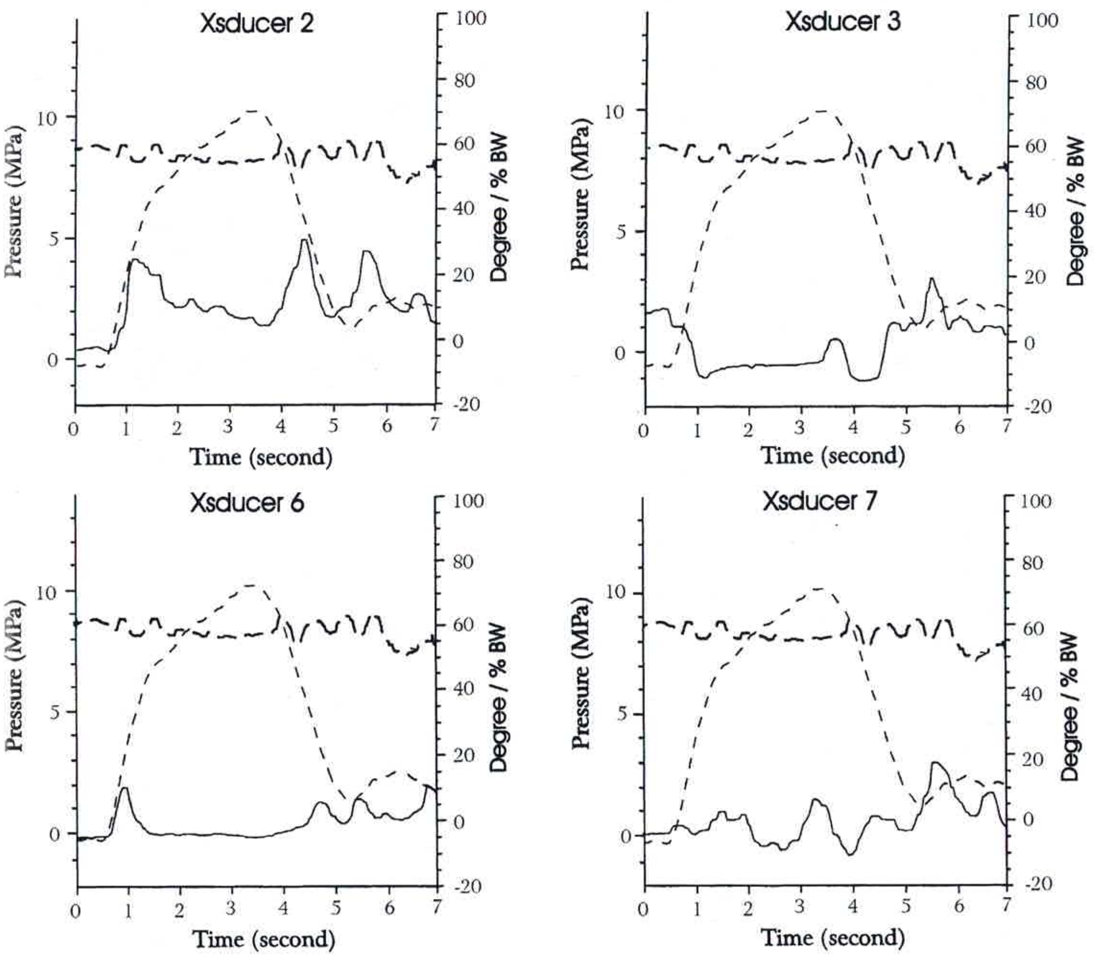

Fig. 4. Complete acetabular contact pressure data (left ordinate) from transducers (Xducer number shown above each plot; see Fig. 1 for locations) which measured ⩾1 MPa magnitude change with time (abscissa); and vertical ground reaction force and hip flexion angle (right ordinate) in a back lifting trial. Pressure data are plotted with a solid light line, hip flexion angle shown as a light dashed line and the vertical ground reaction force was plotted with a dark dashed line. Note little overall change in GRF or in hip stress throughout the trial.

DISCUSSION

Leg lifting produced higher acetabular contact pressures than back lifting. These *in vivo* data support the clinical wisdom of suggesting that spinal-impaired patients use leg lifting and maintain the burden close to the body, increasing the load on the hips and thereby presumably reducing spinal loads. Starting with the weight close to the subject generated higher hip pressures than having the weight far away. In leg lifting (Fig. 3) the hip joint orientation was such that the resultant force acts on the antero-superior femoral head region, producing high pressure on transducer No. 7.

In back lifting, knee moments are reported to be very high, whereas the hamstrings' muscle EMG activity was low (37). Because the knee is flexed during leg lifting, the biarticular hamstrings traversing the hip and knee joint are shorter than with the knee straight during back lifting. Bending the knees shortens the lever arm relative to the hip joint, which reduces hamstrings' effectiveness, apparently forcing its power loss to be compensated by increasing contractile force in the intrinsic muscles around the hip joint. This may well explain the increased oxygen consumption during leg lifting, and explain why manual workers resort to back lifting as they become fatigued or the load is increased (20, 34).

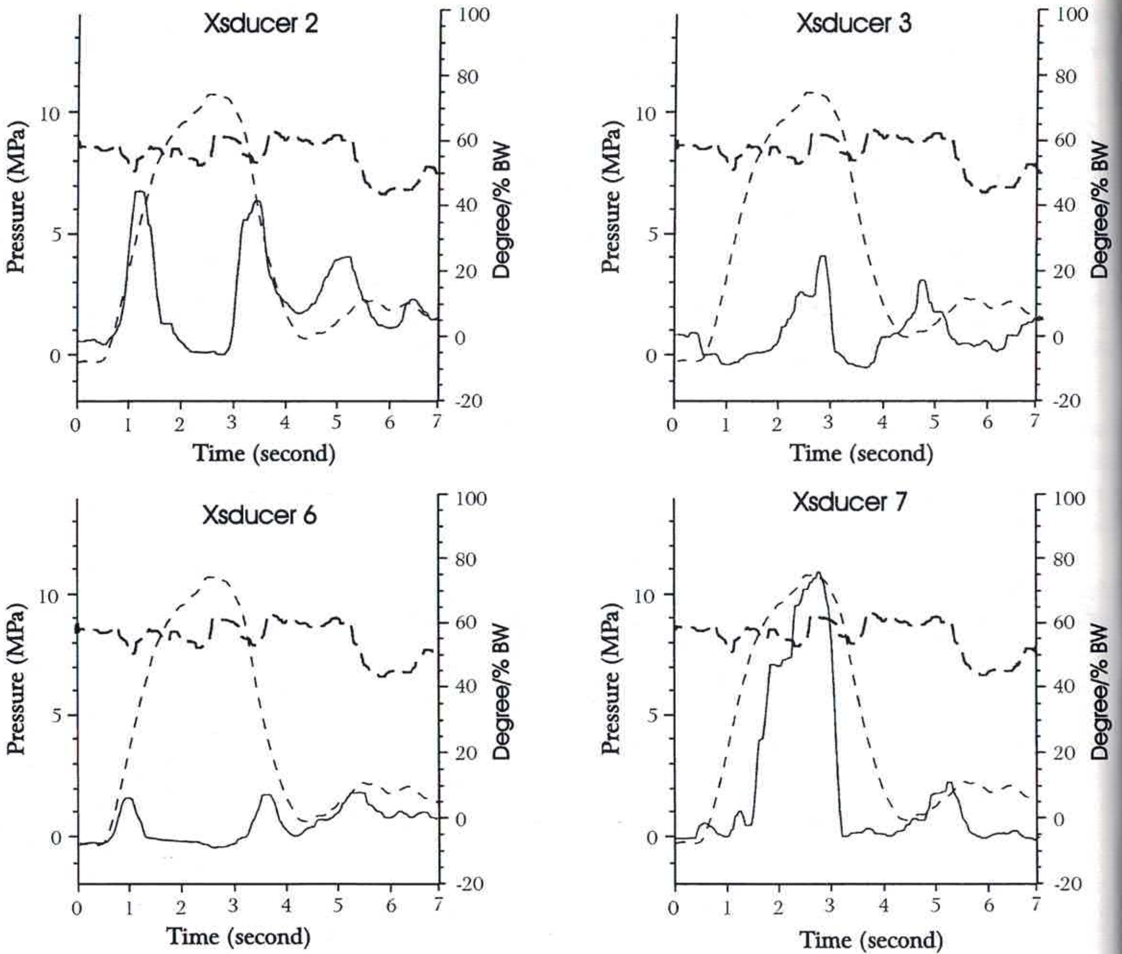

Fig. 5. During a representative leg lifting far trial, peak pressures occurred on transducer 7 simultaneously with maximum hip flexion. Drawing and naming conventions are as described in Fig. 4.

Although hip stabilization by muscle co-contraction is important in leg lifting, hip stability is less important in back lifting. During hip flexion in leg lifting, the hip joint is subjected to high compression forces due to the combination of hip extensor moment and the contribution of ligaments and surrounding tissue (16). In leg lifting, both with the weight far and close, contact pressures at transducer 1 were substantially below 1 MPa, indicating that transducer 1 is not in contact with acetabulum, despite its central location (Fig. 1). Decreasing the prosthesis–cartilage contact area could explain why a higher peak pressure is recorded in leg lifting with the weight close to the body than in the leg lifting far condition. The smaller total contact area combined with high resultant forces, both from the body weight and hip muscle co-contraction, apparently generated greatest pressure when the back was relatively spared from loading during leg lifting.

The different transducers experiencing high pressure in back lifting compared with leg lifting might result from the proximity on the acetabulum of the transducers reporting the pressure distribution. That is, small rotations in femuro-pelvis orientation apparently redistribute resultant hip forces to focus at different acetabular locations. In addition, the anatomical structure of the acetabulum includes the acetabular fossa depression in the middle of the acetabular dome (Figs. 6 and 7). This changing contact area during different lifting strategies probably explains the low contact pressures during back lifting during experiment 1 (Table I). Any transducers not in contact with acetabular cartilage will report very low or zero pressure magnitudes (21).

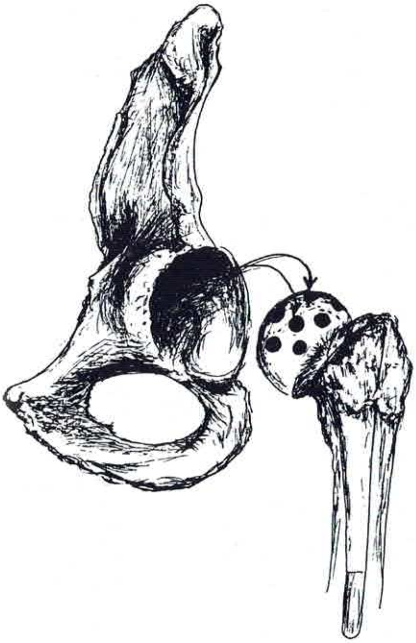

Fig. 6. The acetabular fossa and transducer orientations may permit irregular contact because the acetabulum is not a perfect concave mate for any femoral head: only a portion of the femoral head can be on the acetabular cartilage weight bearing surface at any time.

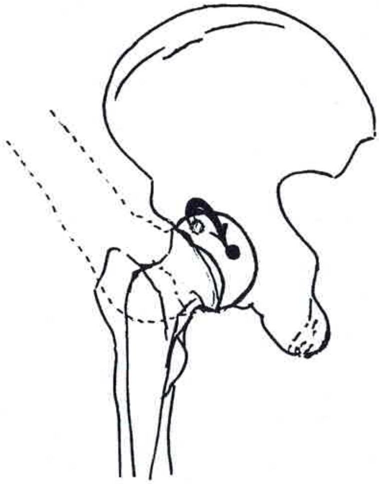

Fig. 7. Femoral head and acetabulum orientations during hip flexion. Because transducer 7 is rotated backward when the hip is flexed, the contact area with the acetabulum was shifted from anterio-superior to posterio-superior quadrant (see Fig. 3A–D).

During leg lifting, the highest pressure occurred on transducer 7 at maximum hip flexion. Transducer 7 is located on the antero-superior aspect with the hip in the anatomical position (Fig. 1), but it contacts the postero-superior aspect of acetabulum when the hip flexes (Fig. 7), the orientation through which the joint reaction force passes during full hip flexion. Degenerative changes in the articular surface of the acetabulum occur primarily on the postero-superior aspect (8, 13), corresponding to the locations of peak contact pressures in the present *in vivo* data. Thus leg lifting puts more stress on the postero-superior region, perhaps contributing to more hip cartilage degeneration, than does back lifting (7, 35, 38). Furthermore, peak acetabular contact pressures during gait are typically 4–6 MPa (21, 25). Therefore, our data suggest that clinicians should advise person who have traumatic or degenerative hip disease to avoid performing repeated leg lifting.

Limitations of this study include the proviso that the subject is older than most manual workers, and data were collected on a single subject thus decreasing the credibility of generalizations that can be made from these experiments. However, they are the only *in vivo* direct measurements made to date on the effect of different lifting approaches on the hip.

CONCLUSIONS

These data describe a mechanism whereby leg lifting may indeed protect the back, sparing low back structures from excess loading, but at the expense of substantially greater hip cartilage stress. The articulating surfaces where degenerative acetabular cartilage changes occur (8, 13, 35) match quite well the *in vivo* peak pressure locations from the present data. Thus, repeated occupational-related lifting using the "leg lifting" technique may induce further hip degeneration in persons at risk for hip arthrosis.

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