LOAD MOMENTS ABOUT THE HIP AND KNEE JOINTS
DURING ERGOMETER CYCLING

Mats O. Ericson, Åke Bratt, Ralph Nisell, Gunnar Németh and Jan Ekhholm

From the Kinesiology Research Group, Department of Physical Medicine and Rehabilitation and
Department of Anatomy Karolinska Institute, and Department of Mechanics,
Royal Institute of Technology, Stockholm, Sweden

ABSTRACT. The aim of the study was to calculate the
magnitudes of moments of force acting about the bilateral
hip and knee joint axes during ergometer cycling. Six
healthy subjects pedalled a weight-braked bicycle ergo-
meter at different workloads, pedalling rates, saddle
heights and pedal foot position. During cycling at 120
Watts, 60 revolutions per minute with mid-saddle height
and anterior pedal foot position, the mean peak flexing
and extending hip load moments were 34.3 and 8.9 Nm, re-
spectively. Mean peak flexing knee load moment was 28.8 Nm
and extending moment was 11.9 Nm. Hip load moments
were significantly increased by increasing the ergometer
workload or pedalling rate. For knee load moments, work-
load was the most important factor. The flexing knee load
moment did not change with changes in pedalling rate.
Different saddle heights or pedal foot positions had a slight
but not always statistically significant influence on the hip
and knee joint loads. The maximum hip and knee joint load
moments induced during cycling were small compared with
those obtained during other exercises or normal activities
such as level walking, stair climbing, and lifting.

Key words: biomechanics, joint load, physical therapy
exercise, rehabilitation

In the rehabilitation of patients with fragile joint
components, a balance may be sought between the
lowest possible load on these and the need for
efficacious training, e.g. high leg muscular activity
or cardiovascular load. Patients with load-elicited
pain should be able, with appropriate adjustment of
the ergometer cycling, to achieve a minimum of
load on injured or fragile lower limb joint compo-
nents. This would then help them to maintain ade-
quate exercise load, yet avoid any pain increase or
benefit from a reduction of, or relief from, pain.
Biomechanical studies on hip and knee joint mo-
ments induced during exercise on a bicycle ergo-
meter may be useful in the individual optimization
of cycling exercise and for comparing ergometer
cycling with other therapeutic exercizes (19, 25), or
with everyday activities such as level walking (2,
17), stair climbing (1) or lifting (5, 18).

The bicycle ergometer has been used as an exer-
cise apparatus in postoperative care after hip (26),

knee (4, 12) and ankle joint (10, 15) surgery, and in

the rehabilitation of patients with rheumatoit ar-
thritis (24), or various kinds of knee dysfunction
(13, 14, 16).

Several studies of joint load during cycling have
been presented recently (6, 8, 10, 12, 27). Van
Elegem et al. (27) performed a vector analysis of
knee joint forces about the bilateral knee joint axis
(about which flexion and extension motions occur)
during cycling racing. They estimated a compres-
sive knee joint force of 75 kg (corresponding to 736
N) for a person weighing 60 kg. Henning et al. (12)
performed an in vivo strain gauge study of elonga-
tion of the anterior cruciate ligament during differ-
ent activities. They found that cycling produced
only 7% of the elongation occurring in an 80 pound
(365 N) Lachman test, and proposed that the proper
order for a rehabilitation program should be crutch
walking, cycling, walking, slow running, and faster
running. Ericson et al. (6) determined the knee joint
load moment about the antero-posterior knee joint
axis (about which varus and valgus moments occur)
during exercise on a bicycle ergometer and found
that the load was about the same as induced during
level walking. In another study (8), the dorsiflexing
load moment about the bilateral axis of the talo-
crural joint was reported to be 30.9 Nm. Recently
Gregor et al. (9) reported on the hip, knee and ankle
joint moments induced during cycling. They found
that there was an extending knee load moment
(knee flexor moment) starting approximately half-
way through the propulsive phase of crank rotation
(0-180 degrees). The knee flexor moments were
presented as a creative solution to Lombard's para-
dox (the activity of a two-joint muscle when the
required moment at one of the joints is in the oppo-
site direction to that caused by the muscle).

McLeod & Blackburn (16) used another approach
in a kinematic study where they studied the variation of the inclination of the tibial plateau during cycling and its possible consequences for the knee load. However, neither forces nor loading moments were calculated. The widespread use of cycling in medical rehabilitation supports further biomechanical investigations of cycling as exercise.

The general purpose of the present investigation was to study how hip and knee load moments were affected by changes in ergometer workload, pedaling rate, saddle height and the position of the foot on the pedal.

The following specific questions were asked:

1. What is the magnitude of the load moment about the bilateral hip and knee joint axes during standardized ergometer cycling?
2. How do the maximum hip and knee load moments vary with workload, pedaling rate, saddle height, and position of the foot on the pedal?

**MATERIALS AND METHODS**

Six healthy subjects (who gave informed consent) all men aged between 20 and 31 years (mean = 25.3 years) participated in the study. Their average height and weight were 1.80 m (SD=0.05) and 71.3 kg (SD=5.0). The subjects were students with ordinary daily and recreational cycling experience. None of the subjects suffered from locomotor pain, had previously undergone any joint surgery, or had had any periods of sick leave due to disorders of the musculoskeletal system.

A bicycle ergometer (CardioSecs with weight brakes, Cardiomics, Stockholm, Sweden) with a specially-instrumented pedal (see below) was used. The following variables were used:

1. **Workload (W)**: zero, 120 and 240 Watt (W)
2. **Pedaling rate (rpm)**: 40, 60, 80 and 100 revolutions per minute (rpm)
3. **Saddle height**: low, mid, and high (determined as a percentage: 102, 113, 120% of the distance between the ischial tuberosity and the medially malleolus measured on each subject. The saddle height was measured as the greatest distance from saddle surface to the centre of the upper pedal surface in a straight line along saddle pillar and crank.
4. **Foot position**: one anterior foot position and one posterior foot position were used. The anterior was defined as the position when the centre of the pedal was in contact with the head of metatarsus II (ball of foot), and the posterior foot position approximately 10 cm backward (instep).

The different test combinations studied are shown in Table 1. The reasons for choosing these bicycle parameter values are discussed elsewhere (6). In the present study combination numbers 8 and 10 were excluded from the biomechanical analysis. Cycling in the highest saddle position and the posterior (instep) foot position (comb. 10) gave an unnatural cycling position with tendencies to pelvis rocking and hip motion in the frontal plane. When one of the four variables was changed and studied, the other three were held constant. The one major exception was that the pedaling rate was changed (40, 60, 80 and 100 rpm) a breaking weight of 2 kg was used and hence the workload was 80, 120, 160 and 240 W, respectively. The different workload were regulated by adding weights (zero, 2 and 4 kg) to the weight hooked bicycle ergometer. 120 W, 60 rpm, mid-saddle and anterior foot position were chosen as constant variables. This combination will be referred to as 'standardized ergometer cycling' (comb. 2). In order to eliminate systemic effects of fatigue, the internal sequence of the eleven different test situations was randomized. The saddle heights determined as described above were adjusted to the nearest fixed position with a maximum error of ±1.5 cm. The handlebars were kept level with the saddle. The cyclist's trunk was inclined forward 20-30° from the vertical. All subjects were allowed to warm up and familiarize themselves with the bicycle ergometer. They practised at all the different workloads, pedaling rates, saddle heights and foot positions included in the study.

**Table 1. Summary of the different combinations of the parameters studied**

<table>
<thead>
<tr>
<th>No.</th>
<th>Workload (W)</th>
<th>Pedaling rate (rpm)</th>
<th>Saddle height</th>
<th>Foot position</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0</td>
<td>60</td>
<td>Mid</td>
<td>Anterior</td>
</tr>
<tr>
<td>2</td>
<td>120</td>
<td>60</td>
<td>Mid</td>
<td>Anterior</td>
</tr>
<tr>
<td>3</td>
<td>240</td>
<td>60</td>
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<td>Anterior</td>
</tr>
<tr>
<td>4</td>
<td>80</td>
<td>40</td>
<td>Mid</td>
<td>Anterior</td>
</tr>
<tr>
<td>5</td>
<td>160</td>
<td>60</td>
<td>Mid</td>
<td>Anterior</td>
</tr>
<tr>
<td>6</td>
<td>200</td>
<td>100</td>
<td>Mid</td>
<td>Anterior</td>
</tr>
<tr>
<td>7</td>
<td>120</td>
<td>60</td>
<td>Low</td>
<td>Anterior</td>
</tr>
<tr>
<td>8</td>
<td>120</td>
<td>60</td>
<td>Low</td>
<td>Posterior</td>
</tr>
<tr>
<td>9</td>
<td>120</td>
<td>60</td>
<td>Mid</td>
<td>Posterior</td>
</tr>
<tr>
<td>10</td>
<td>120</td>
<td>60</td>
<td>High</td>
<td>Posterior</td>
</tr>
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<td>120</td>
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<td>High</td>
<td>Anterior</td>
</tr>
</tbody>
</table>

*Combination No. 2 is referred to in the text as 'standardized ergometer cycling'.

All measurements were performed on the left lower limb. In the left pedal, a quartz force transducer (Kistler type 9251 A) was mounted. The equipment allowed forces in the three orthogonal dimensions (x, y and z) to be measured (Fig. 1). In the present study the forces acting in the sagittal plane $F_x$ and $F_z$ was used for calculation of the moment of force acting about the bilateral hip and knee joint axes. The forces were recorded on a UV-recorder (Honeywell 119B Visicorder). A switch was mounted on the bicycle ergometer for marking on the UV-recorder the top position of the crank for each revolution. Time was registered on the UV-recorder parallel to force and crank top position using a specially designed time indication panel with a light-emitting diode display giving a bar representation of time in units down to 1 msec. The different test situations were filmed using a 16 mm ciné-film camera (Pullman Bolex, 60 frames/sec), mounted perpendicular to the sagittal plane of the subject at a distance of 3.5 m. As landmarks for the bilateral hip, knee and ankle joint axes, dye marks were placed on the skin at approximately 1 cm anterior and superior to the tip of the great trochanter of femur, at the centre of the lateral femoral epicondyle and at the tip of the lateral malleolus. Time as indicated by the time indication panel was visible on each film frame.

The subjects cycled for approximately 30 sec on each test occasion before the measurements were taken. A metronome was used to enable each subject to find and keep the correct pedaling rate. The subjects were filmed and the forces registered for 5 sec. One of the approximately five revolutions registered on the UV-recorder was selected and analysed throughout the complete pedal revolution. The film was analysed using an Analyzer (NLA project), which made it possible to 'freeze' the film and trace the picture at intervals of approximately 15 degrees crank angle. This method for registration of lower limb kinematics has been described and used earlier (19, 25). The position for the hip, knee and ankle joint axes were then determined using the traced pictures (Fig. 2). The $F_x$ and $F_z$ force values corresponding to each picture were read from the UV-recorder.

The model used in the present study for calculating hip
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A bicycle ergometer (Cardioonic with weight brakes, Cardiomics, Stockholm, Sweden) with a specially-instrumented pedal (see below) was used. The following variables were studied:
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3. **Saddle height**: low, mid, and high, determined as a percentage (30%, 50%, 70%) of the distance between the ischial tuberosity and the medial malleolus measured on each subject. The saddle height was measured as the greatest distance from saddle surface to the centre of the upper pedal surface in a straight line along saddle pillar and crank.
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When one of the four variables was changed and studied, the other three were held constant. The one major exception was that the pedalling rate was changed (40, 60, and 80 and 100 rpm) a weight breaking weight of 2 kg was used and hence the workload was 80, 120, 160 and 200 W, respectively. The different workload were regulated by adding weights (zero, 2 and 4 kg) to the weight breaking bicycle ergometer. 120 W, 60 rpm, mid-saddle and anterior foot position were chosen as constant variables. This combination will be referred to as 'standardized ergometer cycling' (comb. 2). In order to eliminate systemic effects of fatigue, the internal sequence of the eleven different tests was randomized. The saddle heights determined as described above were adjusted to the nearest fixed position with a maximum error of ±1.5 cm. The handlebars were kept level with the saddle. The cyclist's trunk was inclined forward 25-30° from the vertical. All subjects were allowed to warm up and familiarize themselves with cycling on the specially instrumented bicycle ergometer. They practised at all the different workloads, pedalling rates, saddle heights and foot positions included in the study.

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The model used in the present study for calculating hip
and knee load moments has been described in detail elsewhere (3). It is based upon dynamic mechanics and takes into account the dynamically-induced forces and moment due to forces of inertia and translational motions of the lower limb. With the crank angle, pedal plate angle, joint positions and pedal reaction forces known, the limb motions and joint load moments were calculated. The significance of changes in hip and knee load moments due to changes of workload, pedalling rate, saddle height and foot position was statistically treated and analysed with ANOVA. The level of significance was p<0.05 throughout.

RESULTS

The mean load moments acting about the bilateral hip and knee joint axes during 'standardized' ergometer cycling (120 W, 60 rpm, mid saddle height and anterior foot position) are shown in Fig. 3. Zero and 360° of crank angle correspond to the top position of the pedal and 180° crank angle corresponds to the bottom position. The hip load moment was predominantly flexing, except for the period between approximately 235° and 335° crank angle, where it was extending. The mean peak flexing and extending hip load moments were 34.3 Nm (SD=9.1) and 8.9 Nm (SD=2.6), respectively. The knee load moment was flexing between approximately 100° and 140° crank angle. For the rest of the motion cycle, (140°-300° crank angle) the knee load moment was extending. Mean peak knee flexing and extending load moment were 28.8 Nm (SD=7.5) and 11.9 Nm (SD=2.6), respectively.

The mean peak hip and knee load moments induced at different workloads are shown in Fig. 4. The maximum extending are flexing hip and knee load moments increased with an increase in ergometer workload. The magnitude of the increase tended to be more pronounced for the flexing load moments than for the extending load moments about both the hip and the knee joints.

The mean peak hip and knee load moments induced at different pedalling rates are shown in Fig. 5. An increase in pedalling rate increased the maximum flexing hip load moment and the maximum extending load moments at the hip and knee joints. The maximum flexing knee load moment was not increased with an increased pedalling rate.

The mean peak load moments acting about the hip and knee during cycling at different saddle heights are shown in Fig. 6. Increased saddle height did not significantly alter the maximum flexing hip load moment, but decreased the extending hip load moment. The maximum flexing knee load moment was decreased and the maximum extending knee load moment increased with increased saddle height.

Fig. 4. Mean peak hip and knee joint load moments induced during cycling with three different workloads; 60, 120 and 240 W.

Fig. 5. Mean peak hip and knee joint load moments induced during cycling at four different pedalling rates; 40, 60, 80 and 100 rpm.

Fig. 6. Mean peak hip and knee joint load moments induced during cycling with three different saddle heights; 'low', 'mid', and 'high'.

Fig. 7. Mean peak hip and knee joint load moments induced during cycling with two different pedal foot positions; 'anterior' (ball of foot) or 'posterior' (instep).

Fig. 8. Mean peak hip and knee joint load moments induced during cycling with four different foot positions; 'front', 'middle', 'rear' and 'bottom'.
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The mean peak hip and knee load moments induced during cycling at four different pedal foot positions; anterior (ball of foot) or posterior (instep).
lower than in level walking (2, 17), stair climbing (1), rising exercise with back support (19, 25) or lifting (5, 18). For rehabilitation of patients with coxarthrosis, for example, and tendency to load induced pain, a possible sequence of exercise might thus be to use exercise before normal level walking and stair walking are tried. The extending hip load moment (balanced by hip flexors) during cycling was even lower than the low load moment during level walking and slow jogging. The maximum flexing and extending knee loads induced during "standardized" ergometer cycling were lower than the maximum knee moments obtained during the other activities reported in Table III. For knee patients (e.g. gomorhathrosis or rocamatoid arthritis) cycling can be used early in the rehabilitation process.

When adjusting the bicycle ergometer, or the cycling technique used, in order to change the load moment on a certain joint, the load on another joint may also be changed. This must be considered, particularly when the patient has disorders or pain in more than one joint. Changes in saddle height or pedal foot position alter the geometry of the closed 4-bar linkage (crank, foot, shank and thigh), which alters important physiological and mechanical factors such as muscle length, joint range of motion and moment arms of external forces and muscle forces. Hence, adjustments made to ergometer or technique are important for joint load moments, muscular activity and energy expenditure during cycling. Changes in oxygen consumption (11, 23) and in muscular activity (7) due to changes in saddle height have been described elsewhere.

The hip and knee load moments reported in the present study could, together with calculation of internal forces be further analysed with local biomechanical models to establish, the compressive forces in the hip (20) and knee (21, 22) joints. Such analyses are in progress and ought to extend the possibilities of evaluating the use of ergometer cycling as exercise for different patients with hip and knee joint disorders.

The present study aimed at estimating the magnitude of hip and knee load moments and how these loads can be altered by different adjustment factors. Such information should be clinically valuable when designing rehabilitation programmes for patients' differing needs.

The following summary may provide some guide for clinical practice: 1) The hip and knee load moments induced during ergometer cycling are lower compared with those induced in most other activities. For instance, normal ergometer cycling induces hip extensor load which is about one-third of that in level walking and stair climbing. The load on the knee extensors in ergometer cycling is about half that in stair climbing and one-fifth of that when going downstairs.

2) An increase in ergometer workload increased the hip and knee load moments more than did the other adjustment factors studied. For instance: an increase from zero to 120 W and from 120 to 240 W each approximately doubled the flexing hip and knee loads.

3) There were small changes in hip and knee load moment when cycling at different pedaling rates (40, 80 and 80 rpm), except for the highest pedaling rate studied, 100 rpm, which significantly increased the flexing hip load moment.

4) Increasing the saddle height reduced the flexing knee load moment by one-third.

5) Change in foot position did not alter the hip and knee load moments.

ACKNOWLEDGEMENTS

This study was supported by grants from the Swedish Medical Research Council (5720) and the Kaarina Institute.

REFERENCES

7. Ericson, M. O., Nisell, R., Arboelius, U. & Ek-
Table II. Changes in maximum hip and knee load moments at various adjustments during ergometer cycling, and their statistical significances (p<0.05)

<table>
<thead>
<tr>
<th>Adjustment factor</th>
<th>Flexing hip load moment</th>
<th>Extending hip load moment</th>
<th>Flexing knee load moment</th>
<th>Extending knee load moment</th>
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<tr>
<td>Increased workload</td>
<td>+</td>
<td>+</td>
<td>+</td>
<td>+</td>
</tr>
<tr>
<td>Increased pedalling rate</td>
<td>+</td>
<td>+</td>
<td>+</td>
<td>+</td>
</tr>
<tr>
<td>Increased saddle height</td>
<td>+</td>
<td>+</td>
<td>+</td>
<td>+</td>
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<tr>
<td>Posterior foot position instead of anterior</td>
<td>+</td>
<td>+</td>
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<tr>
<td>= Increased load moment, ns = no significant change in load moment</td>
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<td>= Decreased load moment</td>
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</tbody>
</table>

The mean peak hip and knee load moments induced with different pedal foot positions are shown in Fig. 7. The use of the posterior foot position instead of the anterior did not significantly alter the hip and knee load moments.

A summary of the changes in hip and knee load moments obtained during alterations of the parameters studied is given in Table II.

DISCUSSION

The method used for estimating the hip and knee load moments induced during cycling was based upon measurement of the forces applied to the pedal on cine-film recordings of lower limb kinematics and on a dynamic mechanical calculation model.

In an earlier study (3) we showed that the pedal plane angle (P) (Fig. 2) was a very sensitive error factor in calculation of the magnitude of hip and knee load moments. An error in measuring or recording the pedal plane angle could cause an error in the direction of the resultant pedal reaction force. This in turn would introduce errors in the mechanical calculation. We determined a maximum test-retest error of ±1° for the pedal plane angle that introducing errors of up to 7.5 and 8.6% in the calculation of the peak hip and knee load moments respectively. The temporal patterns and magnitude of hip and knee moments were in general agreement with those recently reported by Gregor et al. (9). Our calculation model includes all dynamic mechanical components such as forces of inertia, which are important when calculating joint moments during fast motions such as cycling. It might be useful to compare the hip and knee load moments obtained during ergometer cycling with the load induced during other therapeutic exercises or common activities (Table III). The maximum flexing hip load moment (balanced by the hip extensors) induced during ergometer cycling was somewhat higher than during slow jogging (28) but lower than in level walking (2, 17), stair climbing (1), rising exercise with back supported (19, 25) or lifting (5, 18). For rehabilitation of patients with coxarthrosis, for example, and tendency to load induced pain, a possible sequence of exercise might be to use cycling before normal level walking and stair walking are tried. The extending hip load moment (balanced by hip flexors) during cycling was even lower than the load obtained during level walking and slow jogging. The maximum flexing and extending knee load moments induced during 'standardized' ergometer cycling were lower than the maximum knee moments obtained during the other activities reported in Table III. For knee patients (e.g. gonarthrosis or rheumatoid arthritis) cycling can be used early in the rehabilitation process.

When adjusting the bicycle ergometer, or the cycling technique used, in order to change the load moment on a certain joint, the load on another joint may also be changed. This must be considered, particularly when the patient has disorders or pain in more than one joint. Changes in saddle height or pedal position alter the geometry of the closed 4-bar linkage (crank, foot, shank and thigh), which alters important physiological and mechanical factors such as muscle length, joint range of motion and moment arms of external forces and muscle forces. Hence, adjustments made to ergometer or technique are important for joint load moments, muscular activity and energy expenditure during cycling. Changes in oxygen consumption (11, 23) and in muscular activity (7) due to changes in saddle height have been described elsewhere.

The hip and knee load moments reported in the present study could, together with calculation of internal forces be further analyzed with local biomechanical models to establish, the compressive forces in the hip (20) and knee (21, 22, 23) joints. Such analyses are in progress and ought to extend the possibilities of evaluating the use of ergometer cycling as exercise for different patients with hip and knee joint disorders.

The present study aimed at estimating the magnitude of hip and knee load moments and how these loads can be altered by different adjustment factors. Such information should be clinically valuable when designing rehabilitation programmes for patients' differing needs.

The following summary may provide some guide for clinical practice:

1) The hip and knee load moments induced during ergometer cycling are low compared with those induced in most other activities. For instance, normal ergometer cycling induces hip extensor load which is about one-third of that in level walking and stair climbing. The load on the knee extensors in ergometer cycling is about half that in stair climbing and one-fifth of that when going downstairs.

2) An increase in ergometer workload increased the hip and knee load moments more than did the other adjustment factors studied. For instance: an increase from zero to 120 W and from 120 to 240 W each approximately doubled the flexing hip and knee loads.

3) There were small changes in hip and knee load moment when cycling at different pedalling rates (40, 60 and 80 rpm), except for the highest pedalling rate studied, 100 rpm, which significantly increased the flexing hip load moment.

4) Increasing the saddle height reduced the flexing knee load moment by one-third.

5) Change in foot position did not alter the hip and knee load moments.

ACKNOWLEDGEMENTS

This study was supported by grants from the Swedish Medical Research Council (5728) and the Karolinska Institute.

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MEASUREMENT OF SKIN MOBILITY IN THE UPPER BACK

Gerald G. Hirschberg, Irving Fatt and Rebecca Dinn Brown

From the Departments of Physical Medicine and Rehabilitation, University of California, Irvine, USA and Physiological Optics and Engineering Science, University of California, Berkeley, USA

ABSTRACT. On manual testing for skinfold tenderness greater resistance has been reported in patients with marked skinfold tenderness. On objective measurement of skin mobility, by raising a skinfold with a vacuum pump and by establishing a stress-strain curve, no difference in skin mobility was found between subjects with and without skinfold tenderness. Furthermore, contrary to manual testing, the suction testing causes no pain in subjects with clinical skinfold tenderness. In a second series of suction tests, comparing skin mobility in a subject with relaxed and contracted underlying muscles, it was found that muscular contraction reduces skin mobility by 59%. The conclusion is that resistance felt by manual skinfold testing is not inherent in the structures, but is caused by contraction of underlying muscles because of pain caused by the manual skinfold test.

Key words: skinfold tenderness, panniculosis, fibrositis

BACKGROUND

As early as in 1900, a Swedish physician described a painful rheumatic condition with tenderness of the subcutaneous tissues under the name of panniculitis (7). He attributed the findings to an inflammatory process of the subcutaneous structures. Since that time, many articles have appeared in the European literature repeating the description of the clinical findings but differing in their assessment of pathology (1, 4, 5, 6, 8, 9). The prevailing opinion now is that there is no detectable pathology in the subcutaneous tissues. In the United States, this condition is synonymous with fibrositis, or myofascial pain, two clinical syndromes, which also lack pathological documentation (2, 3).

The clinical diagnosis is made by the skinfold test. This consists of raising a fold of skin with thumb and forefinger (Fig. D). If the test is positive, the patient complains of excruciating pain and the examiner feels a considerable resistance to raising of the skinfold. The resistance has been attributed to pathological adhesion of the skin to deeper structures. Our hypothesis is that the resistance felt may be due to muscular contraction in response to pain.

METHOD OF MEASUREMENT

A bell-shaped chamber with an opening of 3 cm and connected to a hand vacuum pump (Fig. 2) is placed over the skin area to be tested and a skinfold is raised into the chamber by producing a vacuum. The degree of negative pressure which ranges from 0 to 30 cm of mercury is read off the pressure gauge. A dial micrometer gauge, whose stem is in contact with the skin, permits one to read the upward displacement up to 2.5 cm. The raising of the skinfold was painless even for patients with severe skinfold tenderness. Tests were done in the upper dorsal area. The edge of the chamber opening is placed 2 cm to the right or left of the spinous process of D-4 (Fig. 3). Phase I: Comparison of subjects with and without skinfold tenderness

Nine subjects were tested to the right and left of D-4 before and after treatment, for a total of 36 tests. Before treatment, the skinfold tenderness was severe and ade-