Transcutaneous Measurement of the Elastic Properties of the Human Femoral Artery

By David J. Mozersky, M.D., David S. Sumner, M.D., D. Eugene Hokanson, B.S., and D. Eugene Strandness, Jr., M.D.

SUMMARY
Measurements of human femoral arterial wall properties have been obtained transcutaneously in a group of males who were clinically free of arteriosclerosis obliterans. The patients were divided into three groups according to age: under 35, 35–60, and over 60 years. Diameter and changes in diameter were measured using an ultrasonic echo-tracking system. Arterial pulse pressure was determined by the auscultatory method. The results suggest that arteries become stiffer with age. The pressure-strain elastic modulus (Ep) varied from \(2.6 \pm 0.3 \text{ SEM} \times 10^6 \text{ dynes/cm}^2\) in the young group to \(6.3 \pm 1.0 \text{ SEM} \times 10^6 \text{ dynes/cm}^2\) in the old group. These values are similar to those reported by other investigators using invasive and in vitro technics. Although statistical comparisons indicate a significant difference between groups, there was considerable variation within the groups. Not infrequently, values for Ep obtained from individuals in the oldest group fell well within the range of the values obtained in the young group.

Additional Indexing Words: Pressure-strain elastic modulus Ultrasons Echo track Aging

The elastic properties of any material can be described in terms of the degree of deformation (strain) that occurs when force (stress) is applied to the material. In the intact vessel the distending force is the blood pressure and the deformation of the vessel is reflected by the increase in diameter that occurs with each pulse beat. The relationship between pressure and strain can be described in terms of the pressure-strain elastic modulus (Ep) which increases as arterial wall stiffness increases.

The mechanical properties of the human arterial wall have been the subject of numerous investigations. Since most of these studies have been conducted in vitro or under highly artificial in vivo conditions, the information derived from them may not be applicable to the undisturbed vessel in its natural environment. Heretofore no practical technic has been available for evaluating the physical properties of arteries in vivo. Recently, however, ultrasonic echo-tracking systems for recording arterial diameter and wall movement have been developed by Arndt and his colleagues and by Hokanson working in our laboratory. Since these instruments may be used transcutaneously, the motion of the arterial wall can be studied repetitively in vivo.

Atherosclerosis, in its fully developed form, leads to structural changes in all layers of the arterial wall. These anatomic changes alter the viscoelastic responses of the involved vascular wall. As new methods for the treatment of lipid disorders are developed, some of which may be of potential value in the treatment of atherosclerosis, it is imperative that a better knowledge of the natural history of the disease be obtained. Since the echo-tracking device
TRANSUCUTANEOUS MEASUREMENT OF FEMORAL ARTERY

provides a method for studying wall properties noninvasively, it could prove to be a valuable method for detecting subclinical atherosclerosis and for following its progression.

Before pathologic states can be investigated, it is necessary first to determine what changes result from aging alone. Aging alters the histologic structure of the arterial wall, and hence may modify the dynamic response to fluctuations in arterial pressure. The present study of the elastic properties of the common femoral artery in different age groups was undertaken to determine the nature of these changes and to establish baseline values for future studies in patients with atherosclerosis.

Materials and Methods

The dynamic properties of the common femoral artery wall were studied in 36 patients (68 vessels). All the patients selected for this study were judged to be clinically free of arteriosclerosis obliterans by the following criteria: normal femoral, popliteal, and pedal pulses, absence of bruits, and lack of vascular symptoms. In all cases, ankle blood pressure was consistently above brachial pressure and the shape of the digit-volume pulse recorded from the toes was normal. The subjects were subdivided according to age into three groups: (1) the young group included 12 patients (24 arteries) who were less than 35 years old; (2) the middle-aged group, 12 patients (22 arteries) who were between 35 and 60 years of age; and (3) the old group, 12 patients (22 arteries) over 60 years of age.

A modification of the instrument described previously by Hokanson et al.11 was used to evaluate arterial compliance. The new instrument, which permits simultaneous tracking of both the anterior and posterior arterial walls, has recently been described in detail.14 A transducer probe containing a single piezoelectric crystal is placed over the artery and coupled to the skin with an acoustic gel. The crystal is driven to emit repetitive short bursts of 5 MHz sound. As the sound passes through the skin and underlying tissues, a portion of it is reflected from the anterior and posterior arterial walls. These echoes are received by the crystal, and are displayed on an oscilloscope screen. Since the oscilloscope sweep begins with the transmission of the sound burst, the distance of the reflecting interface from the probe is indicated by the position of its echo on the horizontal (time) axis of the oscilloscope screen (fig. 1). Echoes from the vessel walls can be distinguished from those originating from other acoustic interfaces by the fact that they move synchronously with the pulse.

In order to track an echo, an electronic gate is opened for about 0.1 μsec at a selected time interval from the transmission sound burst. This allows one half cycle of the echo at that point to be received. When the arterial wall moves, the phase relationships of the echo are altered. This activates a feedback mechanism which causes the gate to move, thereby preserving the original phase relationships. Thus, the gate locks onto the motion of a single cycle in the echo and follows its movement.

By using two gating circuits the motion of the anterior and posterior arterial walls can be tracked simultaneously. Each gate is positioned at the point where the amplitude of the echo is greatest (fig. 1). The output voltage of the instrument is proportional to the distance between the two gates.

The amplitude of the output of the echo tracking system was constant for equal movements of an echo-reflecting surface up to a rate of 60 repetitions/sec.

Either the anterior or posterior wall may be tracked separately by placing one of the gates on an echo from a stationary acoustic interface and the other on the moving echo from the appropriate wall. The largest echoes are obtained when the transducer is positioned at right angles to the arterial wall. In this position, motion artefacts introduced by acute angles are avoided.

Figure 1

View of oscilloscope screen showing the transmission pulse and anterior and posterior wall echoes. Lower trace is the gating signal.
and the ability to study the anterior and posterior wall at the same cross-sectional and longitudinal plane is insured.

The diastolic diameter of the vessel is obtained by counting the pulses from a 7.8 MHz clock. The counter is on for the elapsed time between the two tracking gates, which are preset on the near and far wall echoes. Since the velocity of sound in tissue is $1.56 \times 10^5$ cm/sec, one μsec is equivalent to 0.78 mm. The number of pulses counted, which is equal to the distance (in tenths of a mm) between the echoes, is displayed digitally by the instrument.

Calibration of the echotracking device is accomplished in a small water tank with the transducer aimed at a sound reflecting platform moved by a micrometer. The system was calibrated after each measurement of wall motion.

The experimental procedure was as follows. Each patient was required to rest supine for 20 min before any studies were undertaken. Systolic pressures in the arms and both ankles were determined and strain gauge plethysmography of both second toes was obtained. After the course of the femoral artery had been outlined by palpation or with the continuous-wave ultrasonic velocity detector, the transducer of the echo track was placed over the artery and manipulated until the maximum echoes from both arterial walls were obtained. No pressure was applied to the skin with the transducer, and acoustic coupling was achieved with a gel. If it was not possible to align the transducer so that both near and far wall echoes could be seen simultaneously, the vessel was not used in the study. Measurement of the distance between the echoes at diastole and recordings of the wall motion were then made. At the completion of the study, three observers independently measured the brachial artery pressure by the auscultatory method.

Calculations

From the studies above, the following information was obtained for each femoral artery: mean diastolic diameter (D), maximum change in diameter with each pulse ($\Delta D$), and arterial pulse pressure ($\Delta P$). These data were used to calculate the strain ($\Delta D/D$) and the pressure-strain modulus of elasticity (Ep). Each calculation was based on the average $\Delta D$ from six to 12 pulses.

The pressure-strain elastic modulus, $Ep$, reflects the circumferential stiffness of the arterial segment. As shown in the formula below, it is a measure of the increase in pressure required to effect a unit increase in circumferential strain. Pulse pressure ($\Delta P$) is expressed in dynes/cm²: $Ep = (D/\Delta D) \times \Delta P$. Mean brachial blood pressure was estimated by adding $\Delta P/3$ to the diastolic pressure.

In all the above calculations, D is considered to be the external diameter. Although there is some uncertainty about the actual part of the wall from which the reflections arise, the errors introduced by this approximation are small. Based on the wall thickness data derived by Learoyd and Taylor, the maximum error is unlikely to exceed 6–10%. Student's t test was used to compare the results of the three age groups.

Results

A typical arterial wall displacement pulse recorded with the echo track is shown in figure 2. Its contour is nearly identical to that of the arterial wall displacement pulse (a) and the arterial pressure pulse (b) obtained from opposite femoral artery.

![Figure 2](http://circ.ahajournals.org/)

Comparison of simultaneously obtained femoral arterial wall displacement pulse (a) and the arterial pressure pulse (b) obtained from opposite femoral artery.

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of the pressure pulse recorded simultaneously from the opposite femoral artery. All the recorded displacement pulses were similar to the one illustrated. The results show several significant differences between the age groups (table 1). Although the mean brachial blood pressure and the systolic pressure did not differ appreciably between the groups, the pulse pressure of the old group was significantly higher than the pulse pressure of the young group ($P < 0.02$). The slight increase in average arterial diameter observed in the old group was primarily related to the inclusion in that group of two arteries with diameters of 14 mm and 18 mm.

With increasing age, the average strain ($\Delta D/D$) decreased from 0.026 to 0.017. The differences in strain were significant between the young and middle-aged groups ($P < 0.02$), and the young and old groups ($P < 0.005$), but were not significant when the old and middle-aged group were compared.

The $Ep$ showed significant differences between all groups. The young vessels had significantly lower values than both the middle-aged ($P < 0.02$) and the old groups ($P < 0.005$). The $Ep$ of the middle-aged vessels was significantly lower than that of the old group ($P < 0.05$). These results indicate a progressive increase in stiffness of the femoral artery with age.

The distribution of the values for increase in diameter ($\Delta D$) and $Ep$ are shown in figures 3 and 4. It can be seen that there was considerable variation within the individual groups and that this variation also tended to increase with age. Although it is evident that the trends coincide with those of the average values shown in table 1, several of the old arteries were just as compliant as many of the arteries in the young group.

The pressure-strain elastic modulus ($Ep$) of the left and right femoral arteries in the same individual usually differed somewhat. In the young group, this difference averaged $1.2 \pm 0.9 \times 10^6$ dynes/cm²; in the middle age group, $1.5 \pm 0.9 \times 10^6$ dynes/cm²; and in the old group $1.9 \pm 0.6 \times 10^6$ dynes/cm². Although there was more variability with increasing age, the differences between groups were not statistically significant.

With the transducer held mechanically, serial measurements of the $\Delta D$ will vary depending upon changes in pulse pressure, mean pressure, and the vascular tone. For example, in one young individual, the average $\Delta D$ over a 10-min period was 0.37 mm with a standard deviation of 0.05 mm. The corresponding $Ep$ was $2.4 \pm 0.4 \times 10^6$ dynes/cm². Reapplication of the transducer would be expected to introduce an additional variation due to unavoidable changes in the orientation of the transducer to the vessel. When the right common femoral artery of a single subject was examined daily for 10 days, the average diameter measurement was 10.7 mm with a standard deviation of 0.7 mm. Serial measurements of the $Ep$ ranged from $1.5 \times 10^6$ dynes/cm² to $3.1 \times 10^6$ dynes/cm², with a mean value of $1.9 \times 10^6$ dynes/cm², and a

Table 1

<table>
<thead>
<tr>
<th>Summary of Stress Strain Data For 68 Common Femoral Arteries</th>
</tr>
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<tbody>
<tr>
<td>Under 35 yrs</td>
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<tr>
<td>---------------</td>
</tr>
<tr>
<td>Arteries studied (no.)</td>
</tr>
<tr>
<td>Systolic pressure (mm Hg)</td>
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<tr>
<td>Mean pressure (mm Hg)</td>
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<tr>
<td>Pulse pressure (mm Hg)</td>
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<tr>
<td>Diameter (D) (mm)</td>
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<tr>
<td>Increase in diameter ($\Delta D$) (mm)</td>
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<tr>
<td>Strain ($\Delta D/D$)</td>
</tr>
<tr>
<td>Pressure-strain elastic modulus ($Ep$) (dynes/cm² × 10⁶)</td>
</tr>
</tbody>
</table>

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standard deviation of $0.5 \times 10^6$ dynes/cm². When each individual value was compared to every other value in order to estimate what the difference between any two random measurements was likely to be, a mean difference of $0.50 \pm 0.47 \times 10^6$ dynes/cm² was found.

Both the anterior and posterior walls of all the femoral arteries studied moved in the same direction with each pulse cycle indicating a general anterior translocation of the artery during systole. The movement of the anterior wall in relation to the transducer greatly exceeded that of the posterior wall so that the diameter of the vessel increased while the artery as a whole translocated (fig. 5a). If pressure was applied to the skin over the artery with the transducer, the arterial diameter decreased significantly in the anterior-posterior plane. Pressure also hindered the anterior translocation of the artery and greatly accentuated the posterior wall motion. With pressure, the posterior wall could be forced to move away from the transducer (fig. 5b).

An important observation was the fact that very moderate pressure on the skin could increase the apparent compliance of the artery significantly (fig. 5b). In one subject with a femoral arterial diameter of 10.9 mm, increasing the pressure reduced the anterior-posterior diameter to 5.5 mm and decreased the $E_p$ from $1.01 \times 10^6$ dynes/cm² to $0.16 \times 10^6$ dynes/cm². Even the slight pressure required to compress the artery to 9.8 mm decreased the $E_p$ to $0.43 \times 10^6$ dynes/cm². Therefore, all the results reported in the present study were obtained using care to avoid any pressure on the skin by merely coupling the transducer to the skin with an acoustic gel.

**Discussion**

The results support the widely held opinion that even in the absence of significant arteriosclerosis, arteries become less distensible with age.³, ⁷, ¹⁶ The wide range of values for the $E_p$ found in each group, but particularly in the old group, clearly indicates that the population was far from homogenous. Although the average figures from the old
Anterior
wall

Posterior wall

Patel et al.

Mozersky

and Learoyd


demonstrated the

properties

of arteries.

Although age.

femoral

arteries.

In vivo

In vitro

Electric caliper

Photoelectrically

Mercury strain gauge

Ultrasound

Ultrasound

Young's modulus of 3 \times 10^6
dynes/cm^2, that of collagen is much higher, 1 \times 10^6
dynes/cm^2. The elastic modulus of smooth muscle varies with its activity but may be as high as 12.7 \times 10^6
dynes/cm^2. Since it is unusual to find an artery, even an atherosclerotic artery, with a circumferential
modulus of elasticity exceeding 60 \times 10^6
dynes/cm^2, it is obvious that the elastic properties of arteries are primarily dependent
upon their content of elastin (and/or smooth muscle). It has been shown that the quantity
of elastin and collagen in the arterial wall vary but little with increasing age. Yet the elastin
fibers tend to fragment and become calcified even in the absence of overt atherosclerosis.
As a result of this fragmentation, more of the load will be sustained by the collagen fibers
leading to an increase in the elastic stiffness of the older arteries.

The values for Ep obtained in this study are similar to those observed by other investigators.
A comparison of the present results with those of other workers can be seen in
table 2. There are apparently few differences between the current results and those derived

Table 2
Pressure-Strain Elastic Modulus (Ep) of Human Femoral and External Iliac Arteries (Comparison of Reported Values)

<table>
<thead>
<tr>
<th>Investigator</th>
<th>Artery</th>
<th>Conditions</th>
<th>Method</th>
<th>Ep ± sd dynes/cm² \times 10^6</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>Age (yrs)</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>&lt;35</td>
</tr>
<tr>
<td>Patel et al.⁴</td>
<td>Femoral</td>
<td>In vivo</td>
<td>Electric caliper</td>
<td>4.3*</td>
</tr>
<tr>
<td>Learoyd and Taylor⁵</td>
<td>Femoral</td>
<td>In vivo</td>
<td>Photoelectrically</td>
<td>4.1</td>
</tr>
<tr>
<td>Schulte et al.⁶</td>
<td>Ext Iliac</td>
<td>In vitro</td>
<td>Mercury strain gauge</td>
<td>0.9 ± 0.2</td>
</tr>
<tr>
<td>Arndt and Kober⁶</td>
<td>Femoral</td>
<td>In vivo</td>
<td>Ultrasound</td>
<td>2.6 ± 1.3</td>
</tr>
<tr>
<td>Mozersky et al.</td>
<td>Femoral</td>
<td>In vivo</td>
<td>Ultrasound</td>
<td>3.3 ± 1.8</td>
</tr>
</tbody>
</table>

*Average age = 33; range = 13–62 years.
†All over 35 years; no upper limit stated.

Abbreviations: Ext Iliac = external iliac artery.
from excised specimens or by invasive means. Arndt and Kober, who used an ultrasonic technic similar to ours, found that the average Ep in the femoral arteries of 12 young men was $0.88 \pm 0.20 \text{ dynes/cm}^2$. In the present report, the average Ep in the young group was $2.64 \pm 1.28 \text{ dynes/cm}^2$. Apparently, the femoral arteries in their study were several times as compliant as those we examined. Their measurements were obtained when the anterior and posterior walls were moving in opposite directions. In the present study, this happened only when pressure was applied over the femoral artery. Moreover, the mean diastolic diameter of their arteries ($7.4 \pm 1.4 \text{ mm}$) was considerably less than the mean diameter of the young arteries in our series ($11.0 \pm 1.4 \text{ mm}$). This fact also suggests that they may have been compressing the artery in the anterior-posterior plane.

As shown in figure 5, when the femoral artery is compressed the apparent compliance of the artery rises significantly. With pressure, the lumen of the femoral artery is converted from a roughly circular to an elliptical shape. The effect of pulse pressure on the ellipse is to expand its minor diameter, that is, its anterior-posterior diameter, rather than to stretch the arterial wall at all points as is necessary when the lumen is circular. Therefore, motion in the anterior-posterior plane is greatly increased and the apparent compliance rises. Because of the sensitivity of the femoral artery to even minor compression, it is necessary to avoid all pressure on the skin if accurate studies of arterial elasticity are to be made. For this reason, the higher strains obtained by Arndt and Kober might be questioned.

The results reported in this study can be criticized on three points. Clearly, the patient population may not have been free of atherosclerosis. Although the subjects all had negative histories and normal peripheral pressures and pulses, it was recognized that, almost certainly, the femoral arteries had some atherosclerotic involvement. Since atherosclerosis is so ubiquitous, it is almost a normal part of the aging process. Fatty streaking and intimal deposits are common before the age of 14 and plaques are frequently present by 29 years of age. Therefore, in the absence of the ability to examine the arterial wall directly, we can only state that the arteries to the limb were free of occlusive arterial disease.

The second criticism involves the method used to estimate the femoral artery pressure. Ideally, pressures should be measured directly from the femoral artery. However, the relative accuracy of the auscultatory method of obtaining systolic and diastolic pressures in the arm has been well documented in the past. Measurements made independently by three observers tend to lower the error. Also, it is well recognized that the systolic and diastolic levels of the blood pressure in the brachial and femoral arteries are quite similar in the absence of occlusive disease. Therefore, we do not feel that using the auscultatory method seriously detracts from the accuracy of the findings. It should also be pointed out that blood pressure measurements were involved only in the calculation of the Ep and compliance, and not in calculating the strain. Moreover, it is possible that the introduction of a needle or catheter into the living femoral artery would alter its viscoelastic responses.

Finally, the method used to measure the femoral arterial diameter is open to question. In order to identify the wall of the artery and to measure its diameter, it was necessary to make some assumptions concerning the location of the reflecting tissue interfaces. It is known that maximum reflections occur when the ultrasonic beam is normal to a surface where there is a difference in acoustic impedance. Based on in vitro experiments conducted in our laboratory, we assumed that the large amplitude echoes were arising from the outer media of the near wall and the inner media of the far wall. Thus, the true outside diameter of the artery should be the calculated distance between the echoes plus one wall thickness. However, it is possible that echoes are being received from other interfaces and that the true external arterial wall diameter is somewhat less (at most a few millimeters) than what we estimated. Because the possible
error is relatively small, the uncertainty regarding the diameter does not significantly alter the results.

Since the arterial wall thickness varies during each pulse cycle from a maximum in diastole to a minimum during systole, ΔD measured from the inner walls would differ slightly from that measured from the outer wall. Based on geometric considerations, uncertainties regarding the component of the vessel wall giving rise to the echoes would introduce errors of only a small percentage.

Two of the most challenging problems in medicine today are atherosclerosis and aging. Noninvasive methods of investigating and following the progression of these two conditions are urgently needed. Since the viscoelastic properties of arteries are known to reflect changes caused by aging and atherosclerosis, the echo-track method appears to be a promising technic for following these changes.

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