Wideband external pulse recording during cuff deflation: a new technique for evaluation of the arterial pressure pulse and measurement of blood pressure

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ABSTRACT Analysis of the external brachial pulse recorded during standard blood pressure cuff deflation with use of a transducer with a wide frequency response has revealed a reproducible pattern with three distinct components that we have labeled K1, K2, and K3. K1 is a low-amplitude, low-frequency signal that is present with cuff pressures above systolic pressure. K2 is a triphasic signal appearing at systolic pressure and disappearing at diastolic pressure, which approximately corresponds to the audible Korotkoff sound. K3 appears with cuff pressure between systolic and diastolic pressure and continues to be present below diastolic pressure. Intra-arterial pressure recordings made with a high-fidelity Millar catheter-tip manometer revealed K2 and K3 analogs. K3 resembles the intra-arterial pressure waveform and when calibrated according to the pulse pressure, noninvasive dK3/dt determinations correlated well with intra-arterial dP/dt measurements. The appearance/disappearance property of K2 was designated as the "K2 algorithm" and represents a new, objective noninvasive method for measurement of blood pressure. The K2 algorithm compares favorably with intra-arterial measurements, is more accurate than the auscultatory technique, and may be especially useful in clinical situations in which the auscultatory technique does not work well.


NONINVASIVE characterization of the arterial pressure pulse has historically been separated into two major areas: contour analysis and determination of peak and minimum values (i.e., blood pressure). Contour analysis requires a graphic recording of the external pulse over a peripheral artery in which the sensor-amplifier system used should have an adequate low-frequency response.1 For blood pressure measurement, the auscultatory technique of Korotkoff has become the most widely accepted noninvasive method, and most of the data about human arterial pressure has been obtained with its use.2

The vibrations appearing on the body surface above the brachial artery and under the distal portion of a blood pressure cuff during standard blood pressure determinations have been the subject of numerous investigations and provide the basis for many automatic blood pressure measurement instruments. The recorded signal is generally filtered with arbitrary cutoffs in an attempt to isolate the audible portion of the signal, i.e., the Korotkoff sound. The signal is usually depicted as being virtually undetectable with cuff pressure above systolic pressure, followed by a series of high-frequency spikes gradually increasing in intensity below systolic pressure, and then diminishing below some arbitrary amplitude threshold, or disappearing altogether, at cuff pressure equal to and below diastolic pressure.

In 1969, Whitcher3 was the first to emphasize that the signal recorded under these circumstances was a function of the frequency response characteristic of the sensor-amplifier used. Despite this, subsequent reports have generally had an inadequate low-frequency response, resulting in signal distortion.4 28

In accordance with the low-frequency response requirement for recording an undistorted external pulse, we recorded signals during blood pressure cuff defla-
tion with use of a sensor-amplifier system with a flat frequency response extending below 0.1 Hz. In this report we describe the nature of these signals, analysis of which has led to the development of a new algorithm for blood pressure measurement and additional information about the arterial pressure pulse.

Materials and methods

**Foil electret sensor (FES).** The external pulse was recorded with a specially designed FES that is similar in principle to conventional electret microphones used for airborne sound reception. The FES is modified so that the diaphragm can withstand static cuff pressures up to 250 mm Hg without a significant change in sensitivity. The FES has high sensitivity and an excellent signal-to-noise ratio. It also has a uniform sensitivity over its entire diaphragm surface, and when coupled to a high-impedance amplifier has a flat frequency response from below 0.1 Hz to above 2000 Hz. In some studies, the external pulse was also recorded with other sensors (e.g., Spacelabs, Inc. piezoelectric microphone) with qualitatively similar observations, indicating that the results reported are not uniquely related to the FES.

**Noninvasive procedures.** External pulses were recorded in 253 subjects (age range from 12 to 90 years) during blood pressure cuff deflations from above systolic to below diastolic pressure in order to characterize the observed signal. Recordings were made with subjects in the supine or sitting position. In all cases the FES was placed over the brachial artery and under the distal portion of a blood pressure cuff.

External pulse recordings were also made simultaneously with auscultatory blood pressure determinations by a trained observer in 51 normobes subjects (nine normotensive and 42 consecutive hypertensive patients at the New York Hospital–Cornell University Medical Center Cardiovascular Center). These recordings were obtained with the subject in the supine position. The pressure in the cuff was controlled by a Hokanson E-10 cuff inflator, and was read both by a mercury column and a Gould-Statham P23 ID pressure transducer connected to a physiologic recording system (Electronic for Medicine/Honeywell VR6). The signals from the FES were coupled into a high-impedance (109 Ω) Keithley electrometer amplifier (model 600B) and then into a VR6 direct-current amplifier.

In these subjects the external pulse was recorded under two conditions: (1) during gradual deflation of a cuff while a physician using a stethoscope marked the onset (phase 1) and disappearance (phase 5) of audible Korotkoff sounds on the trace, and (2) during periods of constant cuff pressure. In the latter case, recordings were made for 10 sec periods, in 5 to 20 mm Hg decrements of pressure, starting from above systolic to below diastolic pressure, and were sampled at 400 Hz by a 12-bit analog-to-digital converter for storage in a Digital Equipment Corporation LSI 11/23 computer. This enabled fast-Fourier transforms to be performed to characterize the spectral components of the recorded signal.

**Invasive procedures.** We also compared the noninvasively recorded external pulse with intra-arterially recorded pressure in 16 subjects (two normal, seven hypertensive, and seven heart failure patients). Intra-arterial pressure was recorded from a brachial artery cannula connected to a fluid-filled Gould-Statham P23 ID pressure transducer in nine subjects, and from a Millar solid-state catheter-tip transducer (model PC-330) in seven subjects. The resonant frequency of the fluid-filled pressure recording system was determined by the pop method and ranged from 6.75 to 9.0 Hz, with a damping factor coefficient of 0.25. Since conventional underdamped fluid-filled systems do not have a sufficient natural resonance frequency to permit an accurate recording of high-frequency intra-arterial phenomena, Millar catheters, which have a nominal natural frequency of 35 kHz, were used to provide optimal blood pressure measurement and to determine whether correlates of the externally recorded pulse could be detected intra-arterially. The tips of both fluid-filled and Millar catheters were positioned 2 cm distal to the blood pressure cuff. The FES was placed over the brachial artery and at the distal end of the blood pressure cuff on one or both arms. In 11 of the studies simultaneous auscultatory determinations of blood pressure were also made in a manner similar to that described in the noninvasive studies.

All subjects gave informed consent under protocols approved by the Committee for Human Rights in Research of Cornell University Medical College.

Results

**Characterization of the external pulse in relation to cuff pressure.** The external pulse changed significantly as a function of cuff pressure, as shown in figure 1. Above systolic pressure a low-frequency signal (energy content mostly below 20 Hz) was present, which we designated K1 (figure 1, a). For each subject, K1 had a characteristic shape, usually with three peaks and two troughs. The second trough represented the separation of the systolic and diastolic portions of K1. The relative amplitudes of the individual peaks and troughs varied greatly between subjects (figure 2). That is, the early systolic peak was larger, smaller, or of the same amplitude as the late systolic peak, and occasionally the early and late systolic peaks merged into one wide systolic wave.

As cuff pressure was slowly reduced and approached systolic pressure, either one or both systolic peaks of K1 increased in amplitude, and at systolic pressure a new component appeared. This component was designated as K2 (figure 1, b). K2 was usually a triphasic signal with an initial negative deflection (K2-a), a rising upstroke (K2-b), and a second negative deflection (K2-c). The frequency spectrum of the external pulse with K2 present contained higher frequencies than K1. A comparison of the spectra in figure 1, a and b, shows there was an increased energy content above 20 Hz when K2 was present.

The timing of the appearance of K2 was related to K1, and the initial K2 was usually associated with the largest K1 systolic peak. This can be seen in figure 2. Occasionally multiple K2s occurred. This is illustrated in figure 2, b and e. Figure 2, b, shows the initial appearance of a double K2 in a patient with aortic regurgitation, with each K2 associated with a K1 systolic peak. A second form of multiple K2 is shown in figure 2, e, where the multiple K2s appear not to be associated with separate K1 systolic peaks, but rather occur in relatively rapid succession.
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As cuff pressure was reduced further, an additional component, K3, developed, usually at about the time of the second K1 peak (figure 1, c). With further cuff deflation, K3 increased in amplitude, merged with K2, and dominated the signal (figure 1, d). This addition of K3 caused the energy content below 20 Hz to increase, and because K2 deflections became sharper, the energy content above 20 Hz increased as well. The domination of the signal by K3 typically caused the second K2 negative deflection (K2-c) to become obscured, and with additional cuff pressure reduction, this part of the signal disappeared (figure 1, e). The continued presence of K2 was reflected by the negative notch formed by K2-a and K2-b (K2-a,b). As cuff pressure was reduced, K3 reached a maximum peak-to-peak amplitude, and then reduced in size.

The time interval between the Q wave of the electrocardiogram and the K2 notch (K2-a,b) shortened with cuff pressure reduction. This is illustrated by the reduction in the Q-K2 interval (Q to K2-a,b) from 305 msec in figure 1, b, to 240 msec in figure 1, e. During this reduction in cuff pressure, the time interval between the Q wave and the onset of K1 (Q-K1) remained essentially constant.
When cuff pressure reached diastolic pressure, K2 disappeared and K1 could no longer be independently observed, leaving K3 (figure 1, f). The loss of these components at diastolic pressure was characterized by the disappearance of the notch K2-a,b. The pure K3 component had a shape similar to an intra-arterial pressure waveform, with almost all the signal's energy below 20 Hz (figure 1, f). Below diastolic pressure, K3 gradually diminished in amplitude, but retained its relative shape for as long as adequate coupling was maintained between the FES and the skin.

The K2 algorithm for the measurement of blood pressure in noninvasive studies. We designated the appearance/disappearance property of K2 as the “K2 algorithm” and formed the hypothesis that the cuff pressures at the onset and disappearance of K2 correspond to systolic and diastolic pressure, respectively. We tested this hypothesis by comparing the visual K2 results with independent simultaneous blood pressure measurements made by the auscultatory technique (figure 3). In our 51 subjects the average pressures were 150 ± 26/76 ± 13 mm Hg (mean ± SD) by the K2 algorithm and 144 ± 26/79 ± 13 mm Hg by auscultation, with close correlations between the two measurements (r = .97, p < .0001 for systolic and r = .96, p < .0001 for diastolic). As can be seen in figure 3, a systematic difference existed between results with the two methods. The K2 algorithm gave higher values for systolic and lower values for diastolic pressure than the auscultatory method. Thus, K2 could be detected visually for a few beats before the Korotkoff sound became audible at systolic pressure, and for a few beats after the Korotkoff sound became inaudible.

Further evidence for the correspondence between K2 and the audible Korotkoff sounds was obtained from four subjects who had auscultatory gaps (i.e., the disappearance and reappearance of sounds midway between systolic and diastolic pressures): in all of them, K2 disappeared and reappeared again during cuff deflation at the same time as audible sound.

Invasive studies. Comparisons of the K2 algorithm with both fluid-filled and Millar catheter intra-arterial measurements are shown in figure 4. K2 determinations of both systolic and diastolic pressure showed significant (p < .0001) correlations with intra-arterial determinations (r = .97). The difference between K2 and intra-arterial determinations of diastolic pressure was not significant, although differences in systolic pressure were (t = 2.78, p < .007). K2 measurements were on average 6 mm Hg lower than intra-arterial determinations of systolic pressure. When the analysis was restricted to measurements from seven subjects in whom intra-arterial pressure was recorded with a Millar catheter, a correlation of r = .99 (p < .0001) was achieved for systolic pressure and a correlation of r = .98 (p < .0001) was noted for diastolic pressure, with

![Figure 3](http://circ.ahajournals.org/) Comparison of K2 algorithm and auscultatory determination of blood pressure (left is systolic, right is diastolic) with accompanying lines of identity.
no statistically significant differences between K2 and Millar methods for any blood pressures.

In 11 of the 16 intra-arterial studies in which simultaneous auscultatory determinations were made, a correlation of \( r = .94 \) (\( p < .0001 \)) was achieved for auscultatory and intra-arterial determinations for systolic and diastolic pressure. Statistically significant differences were found between the two methods for both systolic and diastolic pressures (systolic pressure \( t = 3.87, p < .001 \); diastolic pressure \( t = 3.1, p < .005 \)). Auscultatory measurements were on average 15 mm Hg lower and 6 mm Hg higher than intra-arterial determinations of systolic and diastolic pressure, respectively.

When the analysis was restricted to auscultatory-Millar comparisons (five subjects), a correlation of \( r = .98 \) (\( p < .002 \)) was found for systolic pressure, and a correlation of \( r = .97 \) (\( p < .003 \)) was achieved for diastolic pressure. A statistically significant difference remained for systolic (\( t = 2.38, p < .05 \)) but not for diastolic pressure. Auscultatory measurements were on average 10 mm Hg lower than Millar determinations of systolic pressure.

In the 11 studies in which simultaneous intra-arterial, K2 algorithm, and auscultatory determinations were made, K2-auscultatory results showed excellent correlations for all pressures (systolic pressure \( r = .99, p < .0001 \); diastolic pressure \( r = .96, p < .0001 \)), and in addition, the same systematic bias previously found in the noninvasive study continued to exist. Namely, the K2 algorithm measured a higher systolic and pulse pressure (systolic pressure \( t = 3.9, p < .0015 \); pulse pressure \( t = 5.94, p < .0001 \)) and a lower diastolic pressure (\( t = 3.19, p < .005 \)) than the auscultatory technique, with the differences between the indirect and direct measurement (fluid-filled and Millar) of arterial pressure being significantly less for the K2 than the auscultatory method (systolic pressure \( t = 3.9, p < .002 \); diastolic pressure \( t = 3.19, p < .005 \)).

**Comparison of the external pulse and the intra-arterial pressure waveform.** We investigated the relationship between the components of the external pulse and the high-fidelity Millar recording of the distal intra-arterial pressure. Above systolic pressure, the K1 component was present in the externally recorded signal, but not in the distal intra-arterial recording (figure 5, a). At systolic pressure the intra-arterial recording demonstrated a high-frequency component that occurred at the same time as K2 (figure 5, b). The wide bandwidth recording of the Millar catheter distal to the blood pressure cuff revealed arterial pressure components that occurred concurrently with and displayed the same general characteristics of K2 and K3, respectively (figure 5, c). The distal analog of K2 appeared at a cuff pressure equal to systolic pressure, and disappeared at diastolic pressure (figure 5, b and d).
Figure 5, d, shows the resemblance between the external pulse component K3 (cuff pressure below diastolic) and the intra-arterial pressure. Further evidence for the similarity between the two waves was obtained by comparing the maximum rate of rise in a calibrated K3 (dK3/dt) with the maximum change in arterial pressure (dP/dt). A calibration factor was obtained by relating K3 maximum and the preceding K3 minimum to the systolic and diastolic pressures, respectively, determined noninvasively according to the K2 algorithm. Peak dK3/dt and peak dP/dt were determined by computing and averaging the slope of the steepest visually determined tangent to the upstroke of a calibrated K3 and simultaneously recorded intra-arterial pressure over at least four cardiac cycles. The data from 16 intra-arterial studies (nine fluid-filled, seven Millar) are shown in figure 6 with an accompanying line of identity; dK3/dt was closely correlated to intra-arterial dP/dt (r = .97, p < .001), with no statistically significant difference between the means.

**Discussion**

The results of the current studies support previous reports that most of the energy contained in the external pulse recorded during cuff deflation is below the audible range, and that additional clinical information may be gained by including the nonaudible portions of the signal.

The wideband external pulse wave recorded from the distal portion of the blood pressure cuff changes in a characteristic manner as cuff pressure is reduced from above systolic to below diastolic pressure. The shapes of the waveforms vary from one individual to another, but always remain within the basic framework of the reproducible pattern described in this report.

**Characteristics of K1.** When the cuff pressure is above systolic pressure for the entire cardiac cycle, pulse wave penetration through the occluded artery is incomplete, resulting in a K1 signal. K1 may represent the propagation of a pressure wave as a result of the impact of the arterial pressure pulse against the occluded artery, both through surrounding tissue and through the tightly coupled occluding cuff to the active area of the external pulse sensor. Thus, the shape of K1 may be intrinsically related to the shape of the arterial pressure pulse and may similarly contain clinically relevant information about the physical characteristics of the arterial system. It is noteworthy that all normal subjects exhibited a K1 in which the early systolic peak was larger than the late systolic peak and all subjects who exhibited other variations of K1 were patients with known cardiovascular disease.

**Characteristics of K2.** When the cuff pressure is sufficiently reduced below systolic pressure such that a critical transmural pressure gradient (i.e., intra-arterial minus cuff pressure) is achieved, the pulse wave penetrates to the point where a fluid pathway is achieved across the compressed segment of the artery. K2 may develop as a result of the dynamic events created by the reestablishment of fluid contact between the proximal and distal portions of the occluded artery. As cuff pressure is reduced toward diastolic pressure, the critical transmural pressure gradient may occur earlier in the cardiac cycle, thus explaining the reduced timing interval between the Q wave of the electrocardiogram and the appearance of K2.

Sporadically through the years, investigators have reported an early high-frequency phenomenon or component in waveforms recorded either intra-arterially or...
noninvasively\textsuperscript{3, 19–29} that has been described by a variety of names, including “preanacrotic phenomenon,” “whip,” “initial negative wave (or dip),” and “positive or negative spike (or notch).” Some have even regarded this phenomenon as an artifact.\textsuperscript{30} We suggest these descriptions correspond to K\textsubscript{2}. The mechanism of the origin of Korotkoff sounds has been a subject of debate for many years. It has been attributed to both flow-related phenomena (i.e., turbulence, water hammer, fluid-elastic oscillations in collapsible tubes, etc.) and to pressure-related phenomena leading to the sudden tautening of the partially occluded arterial wall.\textsuperscript{31–33} Our observations, which show that K\textsubscript{2} corresponds to the Korotkoff sound and can be recorded as a pressure phenomenon from within the artery, would favor the latter explanation.

**Measurement of arterial pressure with the K\textsubscript{2} algorithm.** The K\textsubscript{2} algorithm represents a new, objective, and possibly more accurate method for measuring blood pressure.\textsuperscript{34} This is a method based on the visual recognition of a pattern, as opposed to the audible recognition of sound.

Although the auscultatory technique is the most widespread and accepted noninvasive method for blood pressure determination, it has acknowledged difficulties. As Rodbard\textsuperscript{35} previously reported, Korotkoff sound intensity is related to pulsatile blood flow. It is likely that many of the difficulties with the auscultatory technique are associated with diminished or excessive flow. In low-flow, vasoconstricted states such as cardiac failure and shock, sound is often of such low intensity that auscultatory determinations may be unreliable.\textsuperscript{35, 36} In other conditions, such as aortic regurgitation, postexercise and hyperthyroidism, sound is often heard down to a cuff pressure of 0 mm Hg.\textsuperscript{37} Problems also exist in infants, children, obese individuals, in elderly subjects with “pseudohypertension syndrome,” and in individuals with aortic stenosis and with bruits.

In addition to problems associated with the wide fluctuations in sound, there is the continuing theoretical question of which phase of sound is best related to the true diastolic pressure, i.e., muffling or disappearance. Thus, in 1967, the American Heart Association recommended use of phase 4 muffling,\textsuperscript{38} while in 1981
phase 5 disappearance was recommended, except in children.37

Visual recognition of K2 is a more sensitive indication of a negative-to-positive transmural pressure gradient reversal than audible Korotkoff sounds. Thus, in low-flow states K2 may be visually present with diminished audible sound, supporting our observation that blood pressure determination in patients with heart failure is difficult by the auscultatory technique, but not with the K2 algorithm.

Because of its greater sensitivity, the K2 algorithm yields a higher systolic and lower diastolic pressure than the auscultatory technique. This systematic bias was upheld by the intra-arterial measurements that also indicated higher systolic and lower diastolic pressures than the auscultatory technique. These results agree with most previous intra-arterial–auscultatory comparisons.39 The K2 algorithm came closer to matching intra-arterial blood pressure determinations than the auscultatory technique, although there was still a statistically significant difference between fluid-filled intra-arterial and K2 determinations of systolic and pulse pressure. However, when the analysis was restricted to those seven studies in which a solid-state Millar catheter was used, these differences disappeared. Thus, it is possible that use of the K2 algorithm not only results in a more accurate noninvasive determination of blood pressure than the auscultatory technique, but may also be more accurate than an undamped, fluid-filled intra-arterial pressure monitoring system with a low natural resonant frequency.

Automatic blood pressure recorders have been developed that generally operate by filtering certain portions of the signal, usually between 10 and 60 Hz, and then applying arbitrary threshold criteria for the detection of systolic and diastolic pressure. Such recorders have serious limitations, in part because there is no clear cutoff point for either systolic or diastolic pressure, and because the absolute intensity of the sounds may vary greatly both within and between individuals.40 In addition, these measurements usually require that the subject be relaxed and motionless. An automatic method based on pattern recognition of K2 might overcome these limitations.

**K3 and intra-arterial pressure.** The similarity between K3 and the intra-arterial pressure is striking and is supportive of previous studies that have demonstrated a close relationship between external pulse recordings and the intra-arterial pressure pulse.41–43 The shape of K3 can be used to compute an accurate mean arterial pressure, and a calibrated K3 signal might have a number of clinical applications, including the noninvasive determination of the rate of change in arterial pressure.

In conclusion, the external pulse recorded during cuff deflation has been separated into three components. The appearance and disappearance of the second component has led to an objective and more accurate noninvasive method of blood pressure determination. The K2 algorithm compares favorably with invasive measurements and may be particularly applicable in clinical situations in which the auscultatory technique has acknowledged difficulties. In addition, by recording the external pulse during cuff deflation we can combine the two major areas of noninvasive analysis of the arterial pressure (i.e., blood pressure measurement and contour analysis of the pulse) in a single procedure.

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