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(54) Non-invasive method and device of measuring the real-time continuous pressure of fluid in elastic tube and the dynamic compliance of elastic tube

(57) The invention presents a non-invasive method and device of measuring the real-time continuous pressure of fluid fluctuating in an elastic tube and the dynamic compliance of the elastic tube, in which the theory of VLDT (Vascular Loading Decoupling Technique) is used. After searching the initial critical depth and determining the decoupled ratio, a DC controller generates a DC control gain to maintain the elastic tube at critical depth, and an AC controller employs the self-adaptive and Step-Hold control rules to create the pulsation of elastic tube without effect of surrounding tissues, and be capable of measuring the real-time continuous fluid pressure and dynamic compliance of elastic tube.

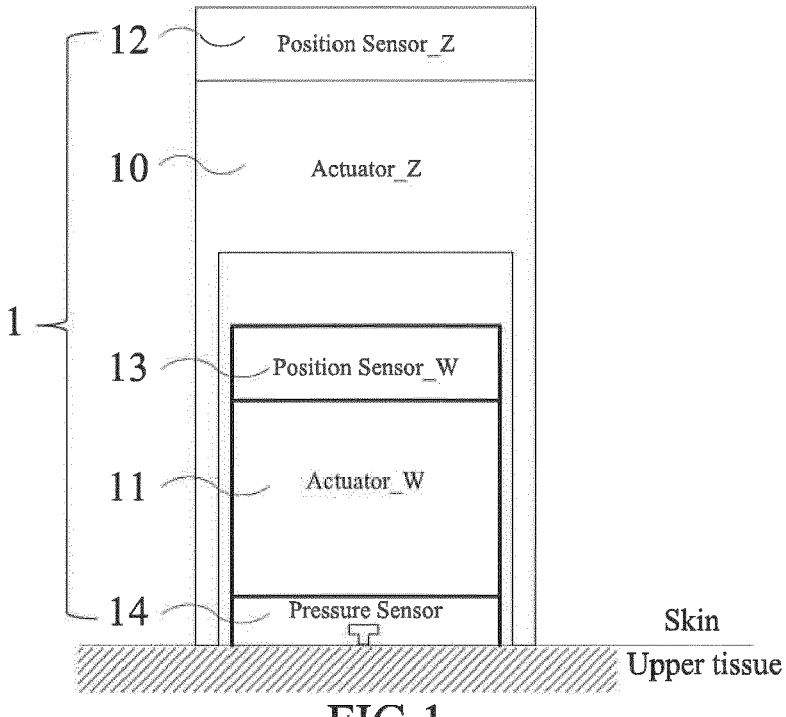

FIG.1

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Description

BACKGROUND OF THE INVENTION

Field of the Invention

[0001] The invention relates to a non-invasive method and device of measuring the pressure of fluid fluctuating in an elastic tube, particularly to a method and device of measuring the actual continuous pressure of fluid inside the elastic tube and the dynamic compliance of the elastic tube.

Description of the Related Art

[0002] The related techniques of the invention are the measurements of the instantaneous blood pressure in artery and the compliance of the blood vessel. The parameters of the blood pressure and arterial compliance are crucial for diagnosis of human health. The present developments are described as follows:

Methods to measure the arterial blood pressure

[0003] Since Marey (1876) first invented the sphygmograph to measure the blood pressure, many researchers have been trying to develop a non-invasive, convenient and reliable instrument to measure the blood pressure in an artery for medicine and health care. In the clinic, blood pressure is measured almost exclusively using non-invasive intermittent techniques, of which the auscultatory and the computerized oscillometric method are most often used. However, both methods only provide a momentary value for systolic blood pressure (SBP), diastolic blood pressure (DBP) and mean blood pressure (MBP).

[0004] Conversely, there are three methods to non-invasively detect the instantaneous arterial blood pressure: arterial tonometry based on the coplanar measurement (Pressman and Newgard 1963, Mackay 1964), Pe˜n'az's method (Pe˜n'az 1973, Wesseling 1984, Wesseling and Pe˜n'az 1986) and the volume-compensation method (Yamakoshi et al 1979, 1980). The latter two are based on the vascular unloading technique (Geddes 1970, Shirer 1962,O'Brien and O'Malley 1991).

[0005] Radial artery tonometry such as HDI CvProfilor (Cohn et al 1995), SphygmoCor (Giot and Dcgautc 1996), Pulsepen (Salv et al 2004) and Colin CBM-7000 arterial tonometry (manufactured by the Colin Corporation, Japan) employs the principle of applanation tonometry, where the artery is partially compressed against a hard structure to provide a continuous read-out of the pulse pressure waveform without the use of an occluding cuff. However, in order to calibrate the pressure measurement, the SBP, DBP and MBP in the opposite arm must also be measured by the auscultatory or cuff-oscillometric method for proportionally estimating the intra-arterial blood pressure. Thus two devices must be used at the same time to measure the blood pressures, which is a major drawback for its medical applications. Besides, the transmission characteristics of the blood pressure from the artery to the skin are not linear or constant.

[0006] Finapres (Boehmer 1987) or Portpres is a non-invasive continuous finger arterial blood pressure monitor based on the vascular unloading technique. With the volume-clamp method of Pe˜n'az, the finger arteries are clamped at a fixed diameter by applying an external pulsating pressure via an inflatable bladder mounted in a finger cuff and a fast-acting servo system. Finapres uses the criteria of Wesseling for determination of the setpoint. The diameter at which the finger arteries are clamped is determined from an infrared plethysmograph mounted in the finger cuff, such that the transmural pressure is zero and the cuff pressure is then equal to the intra-arterial pressure by assuming the ideal transmission of the pulsating pressure from the artery to the cuff. Later, Yamakos et al proposed the volume-compensation method to improve the servo reference, and developed a local pressurization technique to design a pad-type cuff sphygmomanometer (Tanaka et al 2005) for finger and wrist to avoid the occluding cuff encircling the biological segment that makes it uncomfortable in long-term measurements.

[0007] In summary, none of the above methods measures the actual instantaneous blood pressure in an artery, mainly because these methods are not capable of determining the transmission characteristics of the tissue and blood vessel. A possible approach to obtain these characteristics is to decouple the pulsation of the blood vessel from the tissues by using control and identification techniques.

Methods to detect the arterial compliance

[0008] The cardiovascular diseases are mainly caused from arterial angiosclerosis. The estimation of the arterial angiosclerosis is primarily related the compliance of blood vessel. However, it is difficult to directly measure the compliance directly at present, because computing the compliance needs two signals of pressure and the variation of vascular volume (or diameter) simultaneously, and it is not easy to place two sensors to acquire signals at same measuring point.

Hence, the present non-invasive techniques to detect the degree of arterial angiosclerosis mainly involve measuring arterial blood pressure, ABI (Ankle Branch Index), PWV (Pulse Wave Velocity) and AEI (Artery Elasticity Index).

[0009] The most common diagnosis for hypertension disorder is the measurement of systolic blood pressure (SBP) and diastolic blood pressure (DBP) of arterial blood pressure ; ABI (Ankle-Brachial Index) is typically used for assessing the vascular obstruction at lower extremities and suitable for detecting the vascular obstruction caused by thrombus (atherosclerosis); in case of PWV (Pulse Wave Velocity), based on the time reference provided by electrocardiogram (ECG), pulse pressure waveforms of arteries at two electrodes are captured respectively, and the arterial wave velocity for assessing the level of atherosclerosis can be measured via time difference between two pulse waves; and for AEI (Artery Elasticity Index), it uses the continuous blood pressure via modified Windkessel model to compute the Large Artery Elasticity Index (Capacitive Arterial Compliance) C_1 and the Small Artery Elasticity Index (Reflective Arterial Compliance) C_2. It is obvious that the abovementioned measuring techniques are based on continuous blood pressure, not directly detect the arterial compliance. Besides, in order to accurate measuring the compliance of blood vessel, not only to have pressure and variation of vascular volume (or diameter) signals, but also the pulsation of blood vessel should not be affected by surrounding tissue.

[0010] In 2007, the present inventor developed an innovative blood pressure measurement technique, named TCM (Tissue Control Method), which by maintaining the DC part of blood pressure and tracking the AC part of blood pressure to cause the vascular truly unloading; the blood vessel is pulsated without the effect of surrounding tissue. Accordingly, the variation of vascular diameter is obtained, but meanwhile, the AC part of blood pressure is lost, that is the reference pressure for controller is absent as well. For estimating vascular impedance, the self-adaptive control algorithm is adopted, and the peak-to-peak blood pressure of the previous pulse is taken as reference pressure, by which, the beat-based intravascular continuous blood pressure could be obtained, but obviously the value is inaccurate, as shown in FIG. 11; in addition, the obtained arterial vascular impedance is just an approximate mean value in one pulse pressure cycle, as shown in FIG. 10. In other words, although TCM method could lead to the unloading state of the blood vessel, the absence of reference pressure causes the actual continuous blood pressure and the dynamic compliance of the blood vessel to be unobtainable. The present inventor then proposes a real-time based decoupling technique, named VLDT (Vascular Unloading Decoupling Technique), to measure the actual continuous pressure of fluid in the elastic tube and the dynamic compliance of elastic tube as well.

Summary of the Invention

[0011] The present invention is based upon the Vascular Unloading Decoupling Technique (VLDT) to measure the actual continuous pressure of fluid P_b inside the elastic tube and the dynamic compliance of elastic tube C_v in real-time based.

[0012] The principle of the real-time based VLDT is primarily separated the continuous pressure of fluid P_b into two parts, DC part of fluid pressure $\overline{P}_b$ and AC part of fluid pressure ΔP_b. Then control the displacement of DC-driven actuator to maintain the DC part of fluid pressure $\overline{P}_b$ to be the same as the DC part of sensor's pressure $\overline{P}_s$, i.e. $\overline{P}_s = \overline{P}_b$. The pressure sensor is placed on the top of the tissue which is surrounding the elastic tube. This special location is named as critical depth $\overline{\lambda}_s$. At critical depth $\overline{\lambda}_s$, the measured impedance $\overline{Z}_s=\overline{P}_s/\overline{\lambda}_s$ is the impedance of the surrounding tissue $\overline{Z}_t$ only without the impedance of elastic tube involved, i.e. $\overline{Z}_s = \overline{Z}_t = K_t$. It says that the surrounding tissue is decoupled from elastic tube. In the mean time, move the AC-driven actuator to tracking the AC part of fluid pressure ΔP_b to maintain the pressure sensor only have the DC part of fluid pressure $\overline{P}_b$ without AC part of fluid pressure ΔP_b, $P_s = \overline{P}_b$. Then the surrounding tissue will act as the rigid body while the elastic tube is pulsating itself without the effect of the surrounding tissue, that is the impedance of surrounding tissue is unchanged, $Z_t = K_t$; the displacement of AC-driven actuator $\Delta\lambda_s$ is then equal to the variation of elastic tube diameter $\Delta\lambda_v$, i.e. $\Delta\lambda_s = \Delta\lambda_v$. Then the pulsation of elastic tube is decoupled from surrounding tissue as well. This is the key point that we may identify the impedance of elastic tube Z_v under decoupling condition, and in turns to compute the AC part of fluid pressure by $\Delta P_b = -\Delta\lambda_s Z_v$.

[0013] However, after tracking the AC part of fluid pressure ΔP_b completely, the pressure sensor can only have DC part of fluid pressure $\overline{P}_b$ without any AC part of fluid pressure ΔP_b, i.e. $P_s = \overline{P}_s + \Delta P_s = \overline{P}_b$; $\Delta P_s = 0$. Although the variation of the elastic tube diameter is obtained, i.e. $\Delta\lambda_s$, $\Delta\lambda_v$, but without AC part of fluid pressure ΔP_b, it cannot compute the impedance of elastic tube Z_v. To remedy this problem, the present inventor employs the self-adaptive control algorithm which confines the error of AC part of sensor's pressure ΔP_e within the acceptable level by setting the open control loop gain equals to a constant K_a. The open loop gain is the product of AC control gain G_a and the parallel impedance H_1, i.e. G_aH_1, = K_a. For example, if let K_a =199, it will limit the error of AC part of fluid pressure ΔP_b within $1/(1+K_a)$=0.5% of range. Thus, the parallel impedance H_1 will be calculated by K_a times of error of AC part of sensor's pressure divided by the displacement of AC-driven actuator as shown in Fig. 6, i.e. $H_1 = \Delta\tilde{P}_b/\Delta\lambda_s = -K_a\Delta P_e/\Delta\lambda_s$, where the parallel impedance H_1 is the equilibrium impedance of the impedance of surrounding tissue $Z_t = K_t$ and impedance of elastic tube Z_v in

parallel, i.e. $H_1^{-1} = Z_t^{-1} + Z_v^{-1}$. Through the given impedance of surrounding tissue Z_t, the impedance of elastic tube Z_v can then be computed in case that the parallel impedance H_1 is obtained, i.e. $Z_v^{-1} = H_1^{-1} - Z_t^{-1}$. In other word, by adjusting the AC control gain G_a to have a constant open loop gain K_a, it can compute the impedance of elastic tube Z_v and the actual continuous pressure of fluid P_b, provided that the error of AC part of sensor's pressure ΔP_s is given.

[0014] According to the control theory and Fig. 6, the error of AC part of sensor's pressure $\Delta P_s = -\Delta P_e$ is related to the reference pressure $\Delta\hat{P}_b$ as ΔP_s, $= -\Delta\hat{P}_b/(1+K_a)$. In order to have the precise error of AC part of sensor's pressure ΔP_s, obtaining the instant reference pressure $\Delta\hat{P}_b$ is crucial. However, Fig. 6 shows that the AC part of sensor's pressure is the sum of reference pressure $\Delta\hat{P}_b$ and output pressure $\Delta\tilde{P}_b$, of parallel impedance H_1, i.e. $\Delta P_s = \Delta\hat{P}_b + \Delta\tilde{P}_b$. It indicates that the pressure sensor is unable to detect the instant reference pressure $\Delta\hat{P}_b$ at normal control period. The present inventor proposed the Step-Hold control algorithm to estimate the reference pressure at Hold stage and tracking the AC part of fluid pressure ΔP_b at Step stage by turns. At Hold stage (n-1 stage), by stopping the movement of AC-driven actuator, $\Delta\lambda_s(n\text{-}1)=0$, will cause output pressure of parallel impedance is equal to zero, $\Delta\tilde{P}_b(n\text{-}1)=0$, and the AC part of pressure sensor then directly sense the reference pressure signal $\Delta\hat{P}_b(n\text{-}1)$ only, i.e. $\Delta P_s(n\text{-}1)=\Delta\hat{P}_b(n\text{-}1)$. Use three Hold stages at beginning to have three instant reference pressures $\Delta\hat{P}_b(n\text{-}3)$ · $\Delta\hat{P}_b(n\text{-}2)$ · and $\Delta\hat{P}_b(n\text{-}1)$, and use the cubic spline curve fitting technique to estimate the n stage of reference pressure $\Delta\hat{P}_b(n)$, then compute the impedance of elastic tube $Z_v(n)$ and the AC control gain $G_a(n)$ at Hold stage. After that, then goes to the Step stage (n stage) to actuate the AC-driven actuator with AC control gain $G_a(n)$ that maintains the AC part of sensor's pressure is equal to the one of $(1+K_a)$th of reference pressure, i.e. $\Delta P_s(n) = -\Delta\hat{P}_b(n)/(1+K_a)$ and earn the variation of elastic tube diameter from AC part of displacement sensor, i.e. $\Delta\lambda_s(n)\cong\Delta\lambda_v(n)$. Hence the n stage of fluid pressure $P_b(n) = \overline{P}_s(n)-\Delta\lambda_s(n)Z_v(n)$ is attained. After completing the Step stage, return to Hold stage, and repeat the procedure until the measurement is finished. Therefore, VLDT is able to give the real-time continuous fluid pressure $P_b(n)$ and the dynamic impedance of elastic tube $Z_v(n)$ at the stage of n=1,3, 5, ···.

[0015] In summary, one of the objectives of the present invention is to provide a non-invasive method and device of measuring the real-time continuous pressure of fluid inside the elastic tube and the dynamic compliance of the elastic tube for solving the abovementioned problems. By using the Vascular Loading Decoupling Technique (VLDT) to decouple the pulsation of elastic tube from other surrounding tissues, thus the real-time continuous pressure of the fluid fluctuating in the elastic tube and the dynamic impedance of the elastic tube can be accurately measured.

[0016] In case of AC-driven actuator is not capable of fully tracking the AC part of fluid pressure ΔP_b; the present inventor extends the VLDT decoupling theory into the area of partially decoupling of elastic tube. Under these circumstances, it can measure the real-time continuous pressure of fluid in elastic tube as well, but the partially decoupled dynamic impedance of elastic tube is obtained. The detailed descriptions are interpreted in the following related paragraphs.

[0017] Further, the dynamically equivalent mechanical characteristics, such as mass $M_v(n)$, damping $D_v(n)$ and stiffness $K_v(n)$, of the elastic tube can be extracted from the dynamic impedance of elastic tube $Z_v(n)$ by using identification technique, wherein the reciprocal of the stiffness $K_v(n)$ is the dynamic compliance of the elastic tube $C_v(n)$. It should mention that the obtained dynamic compliance of elastic tube is very meaningful, especially used for the measurement of compliance of blood vessel, because the measured compliance of blood vessel has two distinguished features: Firstly, the compliance of blood vessel is measured by the definition of compliance of blood vessel, i.e. $C_v(n) \equiv \Delta\lambda_v(n)/\Delta P_v(n)$, not estimated from the continuous blood pressure by Windkessel model; Secondary, it is measured under the situation of solely pulsation of blood vessel without the effect of other surrounding tissues. Thus the measured compliance is more validity of accuracy than other compliance measuring methods. Therefore, the second objective of the present invention is the measurement of dynamic compliance of elastic tube by using VLDT.

Methods based upon VLDT theory

[0018] The abovementioned elastic tube is wrapped by surrounding tissues, and the fluctuating fluid flows in the elastic tube, wherein the actual continuous fluid pressure P_b and the dynamic compliance of elastic tube C_v are to be measured by a non-invasive method according to the theory of VLDT. The methods based upon VLDT theory are depicted as following steps:

Step 1: Searching initial critical depth $\overline{\lambda}_s$

[0019] At beginning, using Oscillometric Method, press the detection head by DC-driven actuator on the surface of the surrounding tissues which is right above the elastic tube, record the fluid pressure and the displacement of DC-driven actuator until the AC part of fluid pressure ΔP_b is disappeared and then return back to the original depth. Thereafter,

move the detection head to the position called initial critical depth $\overline{\lambda}_s$, where the magnitude of AC part of fluid pressure ΔP_b is maximum. It is also the depth that the DC part of fluid pressure $\overline{P}_b$, to be the same as the DC part of sensor's pressure $\overline{P}_s$, i.e. $\overline{P}_s = \overline{P}_b$.

Step 2: determining the decoupled ratio K

[0020] Hold at initial critical depth $\overline{\lambda}_s$ for few seconds to compute the DC part of fluid pressure $\overline{P}_b$ which is the same of the DC part of sensor's pressure, i.e. $\overline{P}_s = \overline{P}_b$; and to examine the difference of AC part of sensor's pressure $\Delta P_s(\Delta n)$ for each sampling period T_s, where the sampling period is set as 2 milli-seconds. If the maximum of the difference of AC part of sensor's pressure $\Delta P_s(\Delta n)_{\max}$ is exceed of the fully tracking ability of AC-driven actuator for each sampling period $\Delta P_s(\Delta n)_{ref}$, then the partially decoupling of elastic tube is enabled and set the decoupled ratio as $K = \Delta P_s(\Delta n)_{ref}/\Delta P_s(\Delta n)_{\max}$, where $0 < K < 1$. Then set the DC part of sensor's pressure is $\overline{P}_s' = \overline{P}_b + (1-K)\Delta P_b$ and AC part of sensor's pressure is $\Delta P_s' = K\Delta P_b$; If not exceed, decoupled ratio is then set as $K = 1$, it means that the AC-driven actuator is able to fully tracking the AC part of fluid pressure ΔP_b.

Step 3: measuring the impedance of elastic tube $Z_{v2}(n)$ and the fluid pressure $P_b(n)$ in the elastic tube

[0021] The theory of VLDT is to maintain the DC part of sensor's pressure as $\overline{P}_s' = \overline{P}_b + (1-K)\Delta P_b$ and to track the AC part of sensor's pressure as $\Delta P_s' = K\Delta P_b = -\Delta P_e'$ simultaneously.

[0022] Thus, the DC controller employs the self-adaptive control rule to limit the error of DC part of fluid pressure $\overline{P}_b$, within the acceptable level by setting the DC open control loop gain equals to a constant K_d. The DC open loop gain is the product of DC control gain G_d and the parallel impedance H_1, i.e. $G_d H_1 = K_d$. For example, the error of DC part of fluid pressure $\overline{P}_b$ is less than 0.2%, if the DC open loop gain is 499 which is based upon control theory. In other word, the purpose of DC controller is to maintain the press down location to the critical depth $\overline{\lambda}_s'$ (or $\overline{\lambda}_s$, if K=1), where is the location that the DC part of sensor' pressure is equal to the DC part of fluid pressure $\overline{P}_s' = \overline{P}_b + (1-K)\Delta P_b$. In the mean time, the impedance of the surrounding tissues is computed by $Z_t(n) = \overline{P}_s'/\overline{\lambda}_s'$.

[0023] Simultaneously, the AC controller initiates a series of estimation of reference pressure, identification of elastic tube's impedance, tracking the AC part of fluid pressure ΔP_b and computation of fluid during the Step and Hold cycles. The details of AC part of controlling procedure are depicted as follows:

Hold-stage-1

[0024] Use cubic spline curve fitting technique to estimate the reference pressure $\Delta\hat{P}_b'(n)$ according to the previous data $\Delta\hat{P}_b'(n\text{-}3) \cdot \Delta\hat{P}_b'(n\text{-}2) \cdot$ and $\Delta\hat{P}_b'(n\text{-}1)$ which are measured from the AC part of sensor's pressure at AC-driven actuator in idling situation (Hold-stage), i.e. $\Delta\lambda_s' = 0$.

Hold-stage-2

[0025] Identify the parallel impedance $H_1(n)$ and calculate the impedance of elastic tube $Z_{v2}(n)$ by.

$$H_1(n) = -\frac{-K_a \Delta P_s'(n)}{\Delta\lambda_s'(n-1)} = \frac{\Delta P_s'(n) - \Delta\hat{P}_b'(n)}{\Delta\lambda_s'(n-1)} = \frac{-\Delta P_e'(n) - \Delta\hat{P}_b'(n)}{\Delta P_e'(n-1)G_a(n-1)}$$

and

$$\frac{1}{Z_{v2}(n)} = \frac{1}{H_1(n)} - \frac{1}{Z_t(n)}$$

Hold-stage-3

[0026] Calculate the AC control gain $G_a(n) = K_a/H_1(n)$, then enter the Step-stage.

Step-stage-1

[0027] Actuate the AC actuator to tracking the AC part of fluid pressure ΔP_b with control gain $G_a(n)$, and measure the AC part of displacement $\Delta\lambda_s'(n)$.

Step-stage-2

[0028] Compute the fluid pressure $P_b(n)$ by following formula:

$$P_b(n) = \overline{P}_b'(n) + \Delta P_b'(n) = \overline{P}_s'(n) - Z_{v2}(n)\Delta\lambda_s'(n)$$

[0029] Repeat the Hold and Step stages to obtain the impedance of elastic tube $Z_{v2}(n)$ and the fluid pressure $P_b(n)$ at each cycle until the end of the measurement.

Step 4: extracting the dynamic compliance of elastic tube $C_v(n)$ from dynamic impedance of elastic tube $Z_{v2}(n)$

[0030] By using parameter identification technique, a series of dynamic impedance of elastic tube $Z_{v2}(n)$ can provide a series of equivalent mechanical properties such as mass $M_v(n)$, damping $D_v(n)$ and stiffness $K_v(n)$, where the reciprocal of stiffness is the compliance of elastic tube $C_v(n)$.

Device based upon VLDT theory

[0031] To realize the theory of VLDT, the device of the present invention comprises:

a detection unit is arranged on the surfaces of the surrounding tissues, at least consisting of a DC-driven actuator, an AC-driven actuator, a DC displacement sensor, an AC displacement sensor and a pressure sensor, wherein the AC-driven actuator is positioned in the DC-driven actuator and can be moved up and down relative to the DC-driven actuator independently, and the pressure sensor is arranged on the end face of the AC-driven actuator and contacting with the surfaces of the surrounding tissues;
a control unit is used for analyzing and processing signals and electrically connected with two actuators, two displacement sensors and a pressure sensor; Owing to control both of DC part and AC part of fluid pressures, the control unit contains DC controller and AC controller.

[0032] The DC-driven actuator executes a displacement action of pressing the detection unit down to the surface of the surrounding tissues according to the control gain G_d from the DC controller to maintain the DC displacement at critical depth $\overline{\lambda}_s$ all the time, the displacement of the DC-driven actuator is measured by attached DC displacement sensor, and the control gain G_d is processed by the DC controller;

[0033] The AC controller of the control unit generates a signals of AC control gain G_a to drive the AC-driven actuator up and down in order to track the AC part of fluid pressure ΔP_b, and the AC displacement sensor measures the change of the displacement of elastic tube $\Delta\lambda_s$.

[0034] The pressure sensor is used for measuring the pressure P_s, on the surfaces of the surrounding tissues. An analyzing algorithm in the control unit decouples the measured pressure P_s into DC part of sensor's pressure $\overline{P}_s$ and AC part of sensor's pressure ΔP_s feedback to DC and AC controllers for controlling.

Brief Description of the Drawings

[0035]

FIG. 1 is a schematic chart of the measuring device of the present invention.
FIG. 2a is a schematic chart of physical models for the known measurement method (TCM).
FIG. 2b is a schematic diagram of the equivalent impedance model for FIG. 2a.
FIG. 2c is a schematic diagram of the mechanical circuit model for FIG. 2b.
FIG. 2d is a simplified mechanical circuit model for FIG. 2c.
FIG. 3a is a schematic chart of physical models for the present method (VLDT).
FIG. 3b is a schematic diagram of the equivalent impedance model for FIG. 3a.
FIG. 3c is a schematic diagram of the mechanical circuit model for FIG. 3b.
FIG. 3d is a simplified mechanical circuit model for FIG. 3c.
FIG. 3e is a mechanical circuit model under Vascular Loading Decoupling Control.
FIG. 4 is a control block diagram for VLDT which comprises of DC control block and AC control block. The whole control is to track the AC part of arterial blood pressure at critical depth.
FIG. 5 is a DC control block diagram of maintaining the DC part of arterial blood pressure at critical depth.
FIG. 6 is an AC control block diagram of tracking the AC part of arterial blood pressure at critical depth.
FIG. 7a is a measuring procedure for the embodiment of the present invention.
FIG. 7b is the plot for describing the Step 3 of FIG. 7a.
FIG. 8 is a comparison chart of the real and estimated impedances of blood vessel.
FIG. 9 is a comparison chart of the real and estimated arterial blood pressure.
FIG. 10 is a comparison chart of the real and estimated impedances of blood vessel for Tissue Control Method (TCM).
FIG. 11 is a comparison chart of the real and estimated arterial blood pressures for Tissue Control Method (TCM).

Detailed Description of the Preferred Embodiments

[0036] The present invention provides a non-invasive method and device to measure the real-time continuous pressure of fluid fluctuating in the elastic tube, where the elastic tube is surrounded by elastic tissues and the dynamic compliance of the elastic tube as well.

[0037] VLDT theory can be applied to a great variety of elastic tubes as long as the tube body has the elasticity and the fluctuating fluid flows in the tube body. In the embodiments of the present invention, the measurement of the real-time continuous pressure of the arterial blood pressure at wrist and the compliance of the blood vessel is the one of examples. The blood pressure in radial artery surrounded by muscle tissues and skin and the compliance of the radial artery are to be measured.

Key Technique of the Invention

[0038] Three key techniques of the present invention are described as follows:

I. Separating the fluid pressure into two parts and using control techniques to create the decoupled situation to provide the possibility of measuring the impedances of surrounding tissues and elastic tube respectively

[0039] It is quite impossible to measure the fluid pressure in elastic tube directly from the surface of surrounding tissues unless the impedances of surrounding tissues and the elastic tube are known for every measured point. It is also not easy to obtain both impedances of surrounding tissues and the elastic tube while they are moving simultaneously unless the movements of the surrounding tissues and the elastic tube are decoupled, i.e. the surrounding tissues move as a rigid body while the elastic tube is pulsating alone. Therefore, the first key technique of the present invention is to create the decoupled situation for surrounding tissues and elastic tube. The key is to divide the fluid pressure into DC part and AC part. Then a DC controller controls the movement of DC-driven actuator to maintain the DC part of sensor's pressure is equal to the DC part of Fluid pressure $\overline{P}_b$, i.e. the position of critical depth. At critical depth, the impedance of the surrounding tissues can be computed by DC part of sensor's pressure and DC part of displacement. Meanwhile, an AC controller drives the AC-driven actuator to fully track AC part of fluid pressure ΔP_b which causes the AC part of sensor's pressure is zero. Under these circumstances, the movement of surrounding tissues will act as the rigid body, i.e. the impedance of surrounding tissues is unchanged, whiles the elastic tube pulsating itself. It means the movement of surrounding tissues and the elastic tube are decoupled. Then the variation of the elastic tube diameter can be obtained from AC part of displacement on the surface, but the AC part of fluid pressure ΔP_b is lose due to the control theory.

II. Using self-adaptive and Step-Hold control algorithms to estimate reference pressure, identify the impedance of elastic tube, track the AC part of fluid pressure and finally compute the real-time fluid pressure

[0040] In order to remedy this problem, the AC part of fluid pressure ΔP_b will not be fully tracked; a limit of AC part of sensor's pressure within acceptable level is set ; Considering the nonlinear impedances of surrounding tissues and elastic tube, the self-adaptive control rule is used, then the AC open loop gain is set as a constant to ensure the limit of AC part of sensor's within the acceptable level, and the impedance of elastic tube can be calculated from the limit of AC part of sensor's pressure, AC open loop gain an the AC part of displacement. However, the limit of AC part of sensor's pressure is determined by the reference pressure. From Fig. 6, the sum of the reference pressure and the output of parallel impedance are equal to the limit of AC part of sensor's pressure. If the movement of the AC-driven actuator is hold, then the AC part of sensor's pressure will be the same as reference pressure; Thus a Step-Hold control rule as described in the theory of VLDT is adapted to make the measurement of the real time fluid pressure and the impedance of elastic tube to be feasible.

III. Extracting the dynamic compliance of elastic tube without the effect of surrounding tissues from the impedance of elastic tube by using parameter estimation technique

[0041] The impedances of elastic tube contain the dynamic equivalent mass, damping and stiffness that can be further extracted by using parameter identification technique, wherein the reciprocal of the stiffness is the dynamic compliance of the elastic tube, which is the third key technique of this invention. One of the important applications, such as the measurement of the compliance of radial artery is creative and crucial in the diagnosis of angiosclerosis.

Measuring Device

[0042] The measuring device applied to the embodiment of the present invention is shown in the FIG. 1 and FIG. 3a and comprises a detection head 1 and a control unit (which is not shown in the figures), and the detection head 1 is arranged on the surfaces of tissues (the surface of the skin) and comprises a DC-driven actuator 10, an AC-driven actuator 11, a DC part of displacement sensor 12, an AC part of displacement sensor 13 and a pressure sensor 14 in the embodiments of the present invention, wherein the AC-driven actuator 11 is connected with the DC-driven actuator 10 and can independently move up and down relative to the DC-driven actuator 10, the pressure sensor 14 is positioned on the end face of the AC-driven actuator 11 and contacts with the surfaces of the surrounding tissues, the control unit is used for analyzing and processing signals, and the control unit is respectively connected with the DC-driven actuator 10, the AC-driven actuator 11, the DC part of displacement sensor 12, the AC part of displacement sensor 13 and the pressure sensor 14 electrically.

Physical Model for Known Measurement Method (Tissue Control Method, TCM)

[0043] FIG. 2a describes the physical model of the arterial blood pressure at the femur. The point at detection head 1 is labeled n, the skin surface is labeled a, the surface of the femur is d and the points b and c are the blood vessel's upper and lower points respectively. Assuming that the motion model of the arterial blood pressure is lumped, one may express each part, from the detection head 1 to the femur, by a mass element *M*, a damping element *D* and a stiffness element *K* to describe the motion behavior caused by arterial blood pressure transmitted from inside the vessel to the skin above. If the force transmissibility is equivalent to the pressure transmissibility from the blood vessel to the skin, the elements of *M*, *D* and *K* may be defined as

$$M = \frac{P_M}{\ddot{\lambda}}, \quad D = \frac{P_D}{\dot{\lambda}}, \quad K = \frac{P_K}{\lambda} \tag{1}$$

where *P* is the pressure and λ is the displacement. Then, by using the Laplace transform, one may define the impedance Z for each part as

$$Z(s) = \frac{P}{\lambda} = Ms^2 + Ds + K \tag{2}$$

[0044] If the subscripts *h*, *t1*, *v* and *t2* denote the parts of detection head 1, upper tissue, blood vessel and lower tissue, respectively, then the symbols of Z_h, Z_{t1}, Z_v and Z_{t2} indicate the impedances for the corresponding parts of the lumped model.

[0045] If the displacement λ is analogous to the voltage and the pressure P is analogous to the current, then the mechanical impedance will be analogous to the electrical admittance.

[0046] FIG. 2c shows the equivalent mechanical circuit model for the transmission of arterial blood pressure at the femur. One may further combine the impedances of the detection head 1 Z_h, upper tissue Z_{t1}, and the lower tissue Z_{t2} as the impedance of tissue Z_t to simplify the circuit, as shown in FIG. 2d. However, it should be noted that the experimental data show that the impedances are not constants as they vary with the pressed depth.

$$\frac{1}{Z_t} = \frac{1}{Z_h} + \frac{1}{Z_{t1}} + \frac{1}{Z_{t2}}$$

[0047] The abovementioned detection head 1 can be a cuff or an artery tonometry with actuators and sensors. If the artery tonometry is adopted as the detection head 1, the impedance of the detection head 1 is negligible compared with the tissues; however, if the cuff is adopted, the impedance of the detection head 1 should be taken into consideration.

[0048] Based upon the circuit theory, one may express the response of the blood pressure measured on the skin P_s due to excitations of the pressed-down depth λ_s and the arterial blood pressure P_b as

$$P_s = \frac{Z_t Z_v}{Z_t + Z_v}\lambda_s + \frac{Z_t}{Z_t + Z_v}P_b$$

Physical Model for Vascular Loading Decoupliug Technique (VLDT)

[0049] In case of AC-driven actuator 11 is not capable of fully tracking the AC part of fluid pressure ΔP_b; the present inventor extends the decoupling theory into the area of partially decoupling of elastic tube. The physical model is re-constructed as shown in FIG. 3 a.

[0050] The difference between FIG. 3a and FIG. 2a is that FIG. 3a separates the arterial blood vessel into two parts: interior of blood vessel and exterior of blood vessel. The interior of blood vessel is the part of the AC part of blood pressure will be tracking and the exterior of blood vessel is the part of AC part of blood pressure abd combined with DC part of blood pressure will be maintained. Therefore, the equivalent lumped circuit is depicted as FIG. 3c, where the impedances of detection head 1, upper tissue, exterior of blood vessel and lower tissue are in parallel and can be simplified as FIG. 3d. The equivalent impedance of surrounding tissues $\boldsymbol{Z_t}$ is expressed as follows:

$$\frac{1}{Z_t} = \frac{1}{Z_h} + \frac{1}{Z_{t1}} + \frac{1}{Z_{v1}} + \frac{1}{Z_{v2}} + \frac{1}{Z_{t2}} \qquad (3)$$

[0051] And the response of the blood pressure measured on the skin P_s due to excitations of the pressed-down depth λ_s and the arterial blood pressure P_b as

$$P_s = \frac{Z_t Z_{v2}}{Z_t + Z_{v2}}\lambda_s + \frac{Z_t}{Z_t + Z_{v2}}P_b \qquad (4)$$

where the arterial blood pressure P_b can be summed by DC part of blood pressure $\overline{P}_b$ and the AC part of blood pressure ΔP_b.

Determination of decoupled ratio *K* for limiting dilatable range of arterial blood vessel during pulsation

[0052] For each measured point, if the maximum of the difference of AC part of sensor's pressure $\Delta P_s(\Delta n)_{\max}$ is exceed of the fully tracking ability of AC-driven actuator 11 for each sampling period $\Delta P_s(\Delta n)_{ref}$, then the partially decoupling of elastic tube is enabled and set the decoupled ratio as $K = \Delta P_s(\Delta n)_{ref}/\Delta P_s(\Delta n)_{\max}$, where $0<K<1$. Then set the DC part of sensor's pressure is $\overline{P}_s' = \overline{P}_b + (1-K)\Delta P_b$ and AC part of sensor's pressure is $\Delta P_s' = K\Delta P_b$. If not exceed, decoupled ratio is then set as $K = 1$, it means that the AC-driven actuator 11 is able to fully tracking the AC part of fluid pressure ΔP_b.

Analysis of DC part of blood pressure excitation $\overline{P}_b$'

[0053] If one presses the skin to a critical depth $\overline{\lambda}_s$' such that the DC part of sensor's pressure $\overline{P}_s$' measured on the skin equals the DC part of arterial blood pressure $\overline{P}_b$', then equation (4) can be written as

$$\overline{P}_s' = \frac{\overline{Z}_t \overline{Z}_{v2}}{\overline{Z}_t + \overline{Z}_{v2}} \overline{\lambda}_s + \frac{\overline{Z}_t}{\overline{Z}_t + \overline{Z}_{v2}} \overline{P}_b', \qquad \overline{P}_s' = \overline{P}_b' \tag{5}$$

Solving equation (5) gives

$$\overline{P}_s' = \overline{Z}_t \overline{\lambda}_s' = \overline{P}_b' \quad \Rightarrow \overline{Z}_s = \frac{\overline{P}_s'}{\overline{\lambda}_s'} = \overline{Z}_t \tag{6}$$

[0054] Obviously, equation (6) states that at the critical depth $\overline{\lambda}_s$' where, the impedance measured on the skin is equal to the equivalent impedance of detection head 1, upper tissue, exterior of blood vessel and lower tissue only, and the interior of blood vessel is no longer involved in the analysis. It further illustrates that the impedance of the interior of arterial blood vessel, $\overline{Z}_{v2}$ and the impedance of other surrounding tissues, $\overline{Z}_t$ are decoupled.

Analysis of AC part of blood pressure excitation $\Delta\overline{P}_b$'

[0055] When the AC part of arterial blood pressure $\Delta\overline{P}_b$' is imposed on the DC part of the arterial blood pressure $\overline{P}_b$' at the critical depth $\overline{\lambda}_s$', equation (4) can be rewritten as

$$\overline{P}_s' + \Delta P_s' = \frac{Z_t Z_{v2}}{Z_t + Z_{v2}} \left(\overline{\lambda}_s' + \Delta\lambda_s'\right) + \frac{Z_t}{Z_t + Z_{v2}} \left(\overline{P}_b' + \Delta P_b'\right) \tag{7}$$

[0056] Substituting the equation (6) and $\overline{P}_s' = \overline{P}_b' = \overline{Z}_t\overline{\lambda}_s'$ into the equation (7) yields

$$\Delta P_s' = \frac{Z_t Z_{v2}}{Z_t + Z_{v2}} \left(1 - \frac{\overline{Z}_t}{Z_t}\right) \overline{\lambda}_s' + \frac{Z_t Z_{v2}}{Z_t + Z_{v2}} \Delta\lambda_s' + \frac{Z_t}{Z_t + Z_{v2}} \Delta P_b' \tag{8}$$

[0057] If one moves the critical depth $\overline{\lambda}_s$' up and down to keep the AC part of the sensor's pressure measured on the skin $\Delta\overline{P}_s$' equal to zero and maintain the impedance of the tissue $Z_t = \overline{Z}_t$, then equation (8) becomes

$$\Delta\lambda_s' = -\frac{1}{Z_{v2}}\Delta P_b' = \Delta\lambda_v' \tag{9}$$

where $\Delta\lambda_v$' is the variation of the interior of blood vessel diameter. It also represents the displacement varying from the location of the zero transmural pressure of the interior of blood vessel. This derivation reveals that the variation of the displacement measured on the skin is equivalent to the variation of the interior of blood vessel diameter due to the AC part of the arterial blood pressure. The significant implication of this result is that the pulsation of the interior of blood vessel is therefore decoupled from the surrounding tissues, enabling one to identify the impedance of interior of blood vessel and to estimate the intra-arterial blood pressure as shown equation (10) in turn.

$$P_b = \overline{P}_b' + \Delta P_b' = \overline{P}_s' - Z_{v2}\Delta\lambda_s' \tag{10}$$

[0058] For obtaining the situation of (10), we present the LVDT to detect the variation of blood vessel diameter and the real time intra- arterial blood pressure.

Vascular Loading Decoupling Technique (VLDT)

[0059] The key for the VLDT is to track the AC part of arterial blood pressure at critical depth by allowing the blood vessel to pulsate with constant impedance of surrounding tissues. Based upon equations (7) and (8), a control diagram for VLDT is drawn in FIG. 4, where G_d is the block diagram of DC controller and DC-driven actuator 10 and G_a is the AC controller and the AC-driven actuator 11 respectively. Two displacement sensors, DC part of displacement sensor 12 and AC part of displacement sensor 13 are attached with DC-driven actuator 10 and AC-driven actuator 11 and sensing their movements $\overline{\lambda}_s$' and $\Delta\lambda_v$' respectively. Whereas, the pressure P_s measured by the pressure sensor 14 is processed by a FUNC program to obtain a DC part of sensor's pressure $\overline{P}_{ss}$' and an AC part of sensor's pressure ΔP_{ss}', which are taken as feedback signals for DC controller and AC controller, respectively.

System response due to DC part of arterial blood pressure $\overline{P}_b$'

[0060] FIG. 5 is block diagram for DC control of VLDT. The purpose of DC control is to maintain the detection head 1 at critical depth $\overline{\lambda}_s$' where the DC part of sensor's pressure is equal to the DC part of arterial blood pressure, $\overline{P}_s' = \overline{P}_b'$. In order to limit the measuring error of DC part of arterial blood pressure and the nonlinear of impedances of surrounding and blood vessel, the present invention proposes the self-adaptive control rule to set the DC open loop gain K_d, as constant, where $K_d = G_d H_1$. Based upon the DC part of arterial blood pressure is $\overline{P}_s' = \overline{P}_b + K\Delta P_b = \overline{P}_b'$ and control theory, the error of the DC part of arterial blood pressure e_d can be expressed as equation (11).

$$e_d = \frac{1 - H_2}{1 + G_d H_1}\overline{P}_b' \tag{11}$$

where the impedance $H_2 = Z_t / (Z_t + Z_{v2})$ is smaller than 1.

[0061] Therefore, set the DC open loop gain $K_d = G_d H_1 = 499$, the error of DC control results in equation (11) is less than 0.2 %. It means that setting the DC control gain G_d to have DC open loop gain K_d is constant will keep the detection head 1 nearly to the critical depth, which is similar to the analysis of DC part of arterial blood pressure excitation.

System response due to AC part of arterial blood pressure $\Delta\overline{P}_b$'

[0062] When the DC-driven actuator 10 is pressed to reach the critical depth $\overline{\lambda}_s$', an AC controller is needed to move the AC-driven actuator 11 with displacement $\Delta\lambda_s$' in order to have the AC part of arterial blood pressure be zero, $\Delta P_b' = K\Delta P_b = 0$, as shown in FIG. 6.

[0063] The control theory gives the error of the AC part of arterial blood pressure e_a shown in equation (12).

$$e_a = 0 - \Delta P_{ss}' = \frac{-H_2}{1 + G_a H_1} \Delta P_b' \tag{12}$$

[0064] Similarly, the self-adaptive control rule is used. Adjust the AC control gain G_a so as to maintain the AC open loop gain to be the fixed value, $G_a H_1 = K_a$. If set $K_a = 199$, the error of the AC part of arterial blood pressure e_a is smalller than 0.5%. By now, the pulsation of the interior of arterial blood vessel at critical depth is similar to the analysis of AC part of arterial blood pressure $\Delta P_b'$.

[0065] From FIG. 6, it learns that the parallel impedance H_1 can be measured by equation (13).

$$H_1 = \frac{\Delta \tilde{P}_b'}{\Delta \lambda'} = \frac{K_a \Delta P_e'}{\Delta \lambda'} = \frac{-K_a \Delta P_{ss}'}{\Delta \lambda'} \tag{13}$$

[0066] However, the AC part of sensor's pressure is the sum of the reference pressure $\Delta \hat{P}_b'$ and the output pressure $\Delta \tilde{P}_b'$ of parallel impedance H_1, $\Delta P_{ss}' = \Delta \hat{P}_b' + \Delta \tilde{P}_b'$. It cannot be measured with two unknown pressures unless to keep one of the pressure to be zero. Therefore, the present invention provides a Step-Hold control rule to cope with this problem. At Hold stage, let AC-driven actuator 11 is idling, the AC part of displacement is zero and causes the output pressure $\Delta \tilde{P}_b'(n\text{-}1)$ of parallel impedance is zero too, then the measurement of AC part of sensor's pressure is equal to the reference pressure, $\Delta P_{ss}'(n\text{-}1) = \Delta \hat{P}_b'(n\text{-}1)$.

[0067] Use three Hold stages at beginning to have three instant reference pressures $\Delta \hat{P}_b(n\text{-}3) \cdot \Delta \hat{P}_b(n\text{-}2) \cdot$ and $\Delta \hat{P}_b(n\text{-}1)$, and employ the cubic spline curve fitting technique to estimate the n stage of reference pressure $\Delta \hat{P}_b(n)$, then compute the impedance of elastic tube $Z_v(n)$ at Hold stage by equations (14) and (15).

$$H_1(n) = -\frac{-K_a \Delta P_{ss}'(n)}{\Delta \lambda_s'(n-1)} = \frac{\Delta P_{ss}'(n) - \Delta \hat{P}_b'(n)}{\Delta \lambda_s'(n-1)} = \frac{-\Delta P_e'(n) - \Delta \hat{P}_b'(n)}{\Delta P_e'(n-1) G_a(n-1)} \tag{14}$$

$$\frac{1}{Z_{v2}(n)} = \frac{1}{H_1(n)} - \frac{1}{Z_t(n)} \tag{15}$$

and the AC control gain $G_a(n)$ is obtained by equations (16) as well.

$$G_a(n) = \frac{K_a}{H_1(n)} \tag{16}$$

[0068] After that, then goes to the Step stage (n stage) to actuate the AC-driven actuator 11 with AC control gain $G_a(n)$ that maintains the AC part of sensor's pressure is equal to the one of $(1+K_a)$th of reference pressure, i.e. $\Delta P_{ss}'(n) = \Delta \hat{P}_b(n)/(1+K_a)$ and earn variation of elastic tube diameter from AC part of displacement sensor 13, i.e. $\Delta \lambda_s'(n) \cong \Delta \lambda_v'(n)$.

[0069] Finally, compute the n stage of fluid pressure by equation (17).

$$P_b(n) = \overline{P}_s'(n) - \Delta \lambda_s'(n) Z_{v2}'(n) \tag{17}$$

[0070] Repeat the Hold and Step stages to obtain the impedance of elastic tube $Z_{v2}(n)$ and the fluid pressure $P_b(n)$

at each cycle until to the end of the measurement.

Identification of real-time mechanical characteristics (*Mv*, *Dv* and *Kv*) and dynamic compliance *C3* of arterial blood vessel

[0071] In the light of the definition of impedance Z_{v2}, the equivalent mechanical characteristics, such as mass M_v, damping D_v and stiffness K_v constitute equation (18) as follows:

$$-Z_{v2}(s)\Delta\lambda_s'=\Delta P_b' \quad \Rightarrow \quad -(M_v s^2+D_v s+K_v)\Delta\lambda_s'=\Delta P_b' \tag{18}$$

[0072] Matrix parameters of the mass M_v, the damping D_v and the stiffness K_v can be identified via equation (18) through bilinear transform, as shown in equation (19):

$$\begin{bmatrix} M \\ D \\ K \end{bmatrix}=\begin{bmatrix} \Delta\lambda_s'(n) & \Delta\lambda_s'(n-1) & \Delta\lambda_s'(n-2) \\ \Delta\lambda_s'(n-1) & \Delta\lambda_s'(n-2) & \Delta\lambda_s'(n-3) \\ \Delta\lambda_s'(n-2) & \Delta\lambda_s'(n-3) & \Delta\lambda_s'(n-4) \end{bmatrix}^{-1}\begin{bmatrix} Z_{v2}(n)+2Z_{v2}(n-1)+Z_{v2}(n-2) \\ Z_{v2}(n-1)+2Z_{v2}(n-2)+Z_{v2}(n-3) \\ Z_{v2}(n-2)+2Z_{v2}(n-3)+Z_{v2}(n-4) \end{bmatrix} \tag{19}$$

[0073] The mass M_v, the damping D_v and the stiffness K_v of the arterial blood vessel can be further extracted by equation (20) as follows:

$$M_v=\frac{T^2}{16}(M-D+K)\ ;\quad D_v=\frac{T}{4}(M-K)\ ;\quad K_v=\frac{1}{4}(M+D+K) \tag{20}$$

where T is the sampling period.

[0074] The reciprocal of the stiffness K_v is the dynamic compliance of the arterial blood vessel, which is defined as C_3 in the present invention and different from the the Artery Elasticity Indexes of C_1 and C_2, calculated via Windkessel Model based upon the continuous arterial blood pressure. More important, C_3 is measured under the pulsation of the arterial blood vessel without the influence of surrounding tissues.

The Embodiment of Measurement Method of the Invention

[0075] The embodiment of the present invention is shown in FIG. 7a and FIG. 7b, in which the full and the partial tracking AC part of arterial blood pressures are included. The degree of tracking of AC part of arterial blood pressure is determined by the decoupled ratio *K* defined by the capability of AC-driven actuator 11. The measurement of the embodiment mainly comprises the following steps:

Step 1: Searching an initial critical depth $\overline{\lambda}_s$

[0076] At beginning, using Oscillometric Method, press the detection head 1 by DC-driven actuator 10 on the surface of the surrounding tissues to located the initial critical depth $\overline{\lambda}_s$,the DC-driven actuator 10 is right above the elastic tube, record the fluid pressure and the displacement of DC-driven actuator 10 until the AC part of fluid pressure ΔP_b is disappeared and then return back to the original depth. Thereafter, move the detection head 1 to the position called initial critical depth $\overline{\lambda}_s$, where the magnitude of AC part of fluid pressure ΔP_b is maximum. It is also the depth that the DC part of fluid pressure $\overline{P}_b$ to be the same as the DC part of sensor's pressure $\overline{P}_s$, i.e. $\overline{P}_s = \overline{P}_b$.

Step 2: determining the decoupled ratio *K*

[0077] Hold at initial critical depth $\overline{\lambda}_s$ for few seconds to compute the DC part of fluid pressure $\overline{P}_b$ which is the same of the DC part of sensor's pressure $\overline{P}_s$, i.e. $\overline{P}_s = \overline{P}_b$; and to examine the difference of AC part of sensor's pressure $\Delta P_s(\Delta n)$

for each sampling period T_s, where the sampling period is set as 2 milli-seconds. If the maximum of the difference of AC part of sensor's pressure $\Delta P_s(\Delta n)_{max}$ is exceed of the fully tracking ability of AC-driven actuator 11 for each sampling period $\Delta P_s(\Delta n)_{ref}$, then the partially decoupling of elastic tube is enabled and set the decoupled ratio as K = $\Delta P_s(\Delta n)_{ref}/\Delta P_s(\Delta n)_{max}$, where $0 < K < 1$. Then set the DC part of sensor's pressure is $\overline{P_s'} = \overline{P_b} + (1-K)\Delta P_b$ and AC part of sensor's pressure is $\Delta P_s' = K\Delta P_b$; If not exceed, then decoupled ratio is set as $K = 1$, it means that the AC-driven actuator 11 is able to fully tracking the AC part of fluid pressure ΔP_b.

Step 3: measuring the impedance of elastic tube $Z_{v2}(n)$ and the fluid pressure $P_b(n)$ in the elastic tube

[0078] The theory of VLDT is to maintain the DC part of sensor's pressure as $\overline{P_s'} = \overline{P_b} + (1-K)\Delta P_b$ and to track the AC part of sensor's pressure as $\Delta P_s' = K\Delta P_b$ simultaneously.

[0079] Thus, the DC controller employs the self-adaptive control rule to limit the error of DC part of fluid pressure $\overline{P}_b$ within the acceptable level by setting the DC open control loop gain K_d equals to a constant. The DC open loop gain is the product of DC control gain G_d and the parallel impedance H_1, i.e. $G_dH_1 = K_d$. For example, the error of DC part of fluid pressure $\overline{P}_b$ is less than 0.2%, if the DC open loop gain is 499 which is based upon control theory. In other word, the purpose of DC controller is to maintain the press down location to the critical depth $\overline{\lambda_s'}$ (or $\overline{\lambda}_s$, if K=1), where is the location that the DC part of sensor' pressure is equal to the DC part of fluid pressure $\overline{P_s'} = \overline{P_b} + (1-K)\Delta P_b$. In the mean time, the impedance of the surrounding tissues is computed by $Z_t(n) = \overline{P}_s'/\overline{\lambda}_s'$.

[0080] Simultaneously, the AC controller initiates a series of estimation of reference pressure, identification of elastic tube's impedance, tracking the AC part of fluid pressure ΔP_b and computation of fluid during the Step and Hold cycles. The details of AC part of controlling procedure are depicted as follows:

Hold-stage-1

[0081] Use cubic spline curve fitting technique to estimate the reference pressure $\Delta\hat{P}_b'(n)$ according to the previous data $\Delta\hat{P}_b'(n\text{-}3)$ $\Delta\hat{P}_b'(n-3)$ 、 $\Delta\hat{P}_b'(n-2)$ · $\Delta\hat{P}_b'(n\text{-}2)$ · and $\Delta\hat{P}_b'(n\text{-}1)$ which are measured from the AC part of sensor's pressure at AC-driven actuator 11 in idling situation (Hold-stage), i.e. $\Delta\overline{\lambda}_s'=0$.

Hold-stage-2

[0082] Identify the parallel impedance $H_1(n)$ and calculate the impedance of elastic tube $Z_{v2}(n)$ by.

$$H_1(n) = -\frac{-K_a\Delta P_s'(n)}{\Delta\lambda_s'(n-1)} = \frac{\Delta P_s'(n) - \Delta\hat{P}_b'(n)}{\Delta\lambda_s'(n-1)} = \frac{-\Delta P_e'(n) - \Delta\hat{P}_b'(n)}{\Delta P_e'(n-1)G_a(n-1)}$$

and

$$\frac{1}{Z_{v2}(n)} = \frac{1}{H_1(n)} - \frac{1}{Z_t(n)}$$

Hold-stage-3

[0083] Calculate the AC control gain $G_a(n) = K_a/H_1(n)$, then enter the Step-stage.

Step-stage-1

[0084] Actuate the AC actuator to tracking the AC part of fluid pressure ΔP_b with control gain $G_a(n)$, and measure the AC part of displacement $\Delta\lambda_s'(n)$.

Step-stage-2

[0085] Compute the real-time continuous fluid pressure $P_b(n)$ by following equation:

$$P_b(n) = \overline{P}_b'(n) + \Delta P_b'(n) = \overline{P}_s'(n) - Z_{v2}(n)\Delta\lambda_s'(n)$$

[0086] Repeat the Hold and Step stages to obtain the impedance of elastic tube $Z_{v2}(n)$ and the real-time continuous fluid pressure $P_b(n)$ at each cycle until the end of the measurement.

Step 4: extracting the dynamic compliance of elastic tube $C_v(n)$ from dynamic impedance of elastic tube $Z_{v2}(n)$

[0087] By using parameter identification technique, a series of dynamic impedance of elastic tube $Z_{v2}(n)$ can provide a series of equivalent mechanical properties such as mass $M_v(n)$, damping $D_v(n)$ and stiffness $K_v(n)$, where the reciprocal of stiffness is the compliance of elastic tube $C_v(n)$.

Verification

[0088] In order to verify the feasibility of VLDT, nonlinear impedances of surrounding tissues and blood vessel and blood pressures based upon experiments are assumed. The simulation results run by MATLAB are shown in FIG. 8 and FIG. 9. FIG. 8 is a comparison chart of real and estimated of impedances of blood vessel; and FIG. 9 shows the comparison chart of real and estimated of arterial blood pressures. It all indicates that the results are satisfied and feasible in reality.

Conclusion

[0089] From the above, the measurement method and the measuring device provided by the invention are useful for measuring the real time fluid pressure in an elastic tube which surrounded by elastic tissues and the dynamic compliance of elastic tube as well. This invention can be extensively applied for detecting the degree of aging for elastic pipes after long run operations, or the some medical applications which need the real continuous blood pressure or dynamic compliance of blood vessel for more precisely diagnostic analyses.

Claims

1. A non-invasive method of measuring the real-time continuous pressure of fluid in elastic tube and the dynamic compliance of the elastic tube, wherein an elastic surrounding tissues are wrapped outside the elastic tube, a fluctuating fluid flows having a fluid pressure which can be divided into DC part of fluid pressure $\overline{P}_b$ and AC part of fluid pressure ΔP_b in the elastic tube; and the measurement method is characterized as follows:

   Searching an initial critical depth $\overline{\lambda}_s$: using Oscillometric Method, press a detection head (1) by a DC-driven actuator (10) on the surface of the surrounding tissues to locate the initial critical depth.;
   Determining a decoupled ratio K: Hold at initial critical depth to determine the decoupled ratio K for partial or full tracking of the AC part of fluid pressure ΔP_b;
   Updating a critical depth $\overline{\lambda}_s'$: actuate DC-driven actuator (10) to maintain a DC part of sensor's pressure $\overline{P}_s$, is equal to the DC part of fluid pressure $\overline{P}_b$, where is the location of critical depth $\overline{\lambda}_s'$;
   Estimating a reference pressure $\Delta\hat{P}_b'(n)$: set AC-driven actuator (11) in idling situation (Hold-stage) to obtain three previous reference pressures $\Delta\hat{P}_b'(n\text{-}3) \cdot \Delta\hat{P}_b'(n\text{-}2) \cdot$ and $\Delta\hat{P}_b'(n\text{-}1)$, and estimate the reference pressure

$\Delta\hat{P}_b'(n)$ at Step-stage;
Measuring an impedance of elastic tube $Z_{v2}(n)$: use constant AC open loop gain and self-adaptive control rule to estimate the parallel impedance H_1 and compute the impedance of elastic tube $Z_{v2}(n)$;
Calculating an AC control gain G_a: use constant AC open loop gain K_a and parallel impedance H1 to calculate the AC control gain G_a for next Step-stage;
Tracking the AC part of fluid pressure ΔP_b to obtain an AC part of displacement $\Delta\lambda_s'(n)$: send an AC control gain K_a from AC controller, and move the AC-driven actuator (11) to track the AC part of fluid pressure ΔP_b and earn the AC part of displacement $\Delta\lambda_s'(n)$;
Computing a real-time continuous fluid pressure $P_b(n)$ in elastic tube: according to the impedance of elastic tube $Z_{v2}(n)$ and the AC part of displacement $\Delta\lambda_s'(n)$ to compute the AC part of fluid pressure $\Delta P_b'(n)$, then add the DC part of fluid pressure $\overline{P}_b'(n)$, which is equal to a DC part of sensor's pressure $\overline{P}_s$, to have the real-time continuous fluid pressure $P_b(n)$;
Extracting a dynamic compliance of elastic tube $C_v(n)$: a series of equivalent mechanical properties including mass $M_v(n)$, damping $D_v(n)$ and stiffness $K_v(n)$ are extracted from dynamic impedance of elastic tube $Z_{v2}(n)$ by using parameter identification technique. In which the reciprocal of stiffness is the compliance of elastic tube $C_v(n)$.

2. The method of measuring the real-time continuous pressure of fluid in elastic tube and the dynamic compliance of the elastic tube of claim 1, **characterized in that,** the prediction measure is a spline curve fitting technique.

3. The method of measuring the real-time continuous pressure of fluid in elastic tube and the dynamic compliance of the elastic tube of claim 1, **characterized in that**, the method can be further applied to the measurement of a blood pressure in arterial blood vessel, the elastic tube to be measured is the arterial blood vessel and the surrounding tissues are muscle tissues, and the fluid pressure is the blood pressure in the arterial blood vessel.

4. The method of claim 1, wherein the applied measuring device is **characterized in that**:

A detection head (1) arranged on the surface of the surrounding tissues, at least comprises a DC-driven actuator (10), an AC-driven actuator (11), a DC part of displacement sensor (12), an AC part of displacement sensor (13) and a pressure sensor (14), wherein the AC-driven actuator (11) is placed in the DC-driven actuator (10) and can be independently moved up and down relative to the DC-driven actuator (10), and the pressure sensor (14) is positioned on the end face of the AC-driven actuator (11) and contacts with the surfaces of the surrounding tissues;
A control unit is used for analyzing and processing signals and respectively connected with the two actuators (10 and 11), the two displacement sensors (12 and13) and the pressure sensor (14) electrically;
The DC-driven actuator (10) maintaining the DC part of sensor's pressure $\overline{P}_s$ on the surface of the surrounding tissues equals to the DC part of fluid pressure $\overline{P}_b$ in accordance with a DC control gain G_d, which is processed by a DC controller, and the displacement of the DC-driven actuator (10) is measured by the DC part of displacement sensor (12);
The AC-driven actuator (11) is moving up and down along with the AC part of fluid pressure ΔP_b in elastic tube according to the AC control gain G_a, which is processed by an AC controller, and the displacement change $\Delta\lambda_s'$ of the AC-driven actuator (11) is measured by the AC part of displacement sensor (13);
The pressure sensor (14) is used to measure the pressure on the surface of the surrounding tissues P_s while pressing down the DC-driven actuator (10) to maintain the DC part of fluid pressure $\overline{P}_b$ and moving the AC-driven actuator (11) up and down to tracking the AC part of fluid pressure ΔP_b. The measured pressure signals is transmitted to the control unit for signal processing.

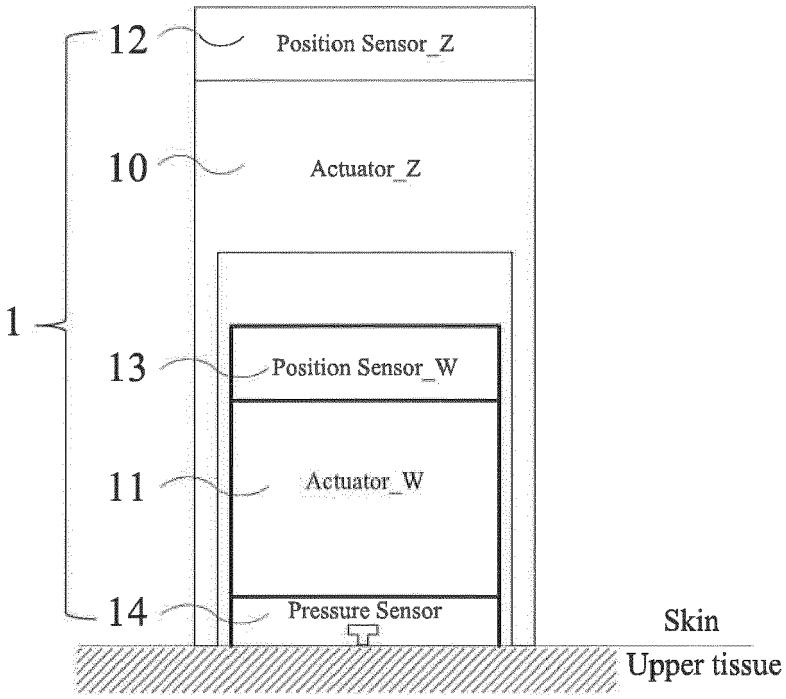
12
Position Sensor_Z
10
Actuator_Z
1
13
Position Sensor_W
11
Actuator_W
14
Pressure Sensor
Skin
Upper tissue

FIG.1

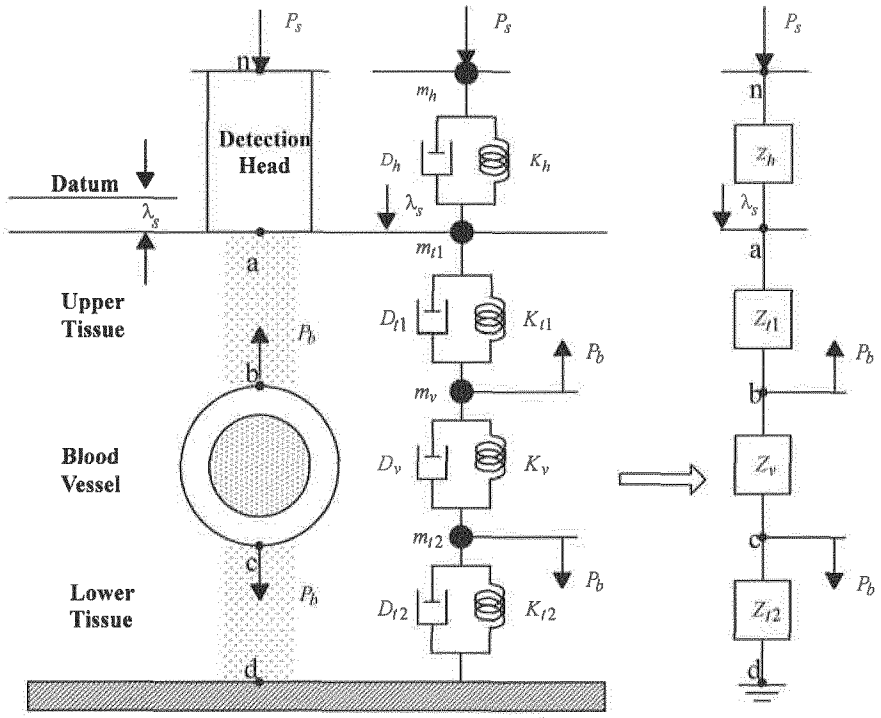
P_s
n
Detection Head
Datum
λ_s
a
Upper Tissue
P_b
b
Blood Vessel
c
Lower Tissue
P_b
d
m_h
D_h
K_h
λ_s
m_{t1}
D_{t1}
K_{t1}
P_b
m_v
D_v
K_v
m_{t2}
P_b
D_{t2}
K_{t2}
Z_h
Z_{t1}
Z_v
Z_{t2}

FIG.2a

FIG.2b

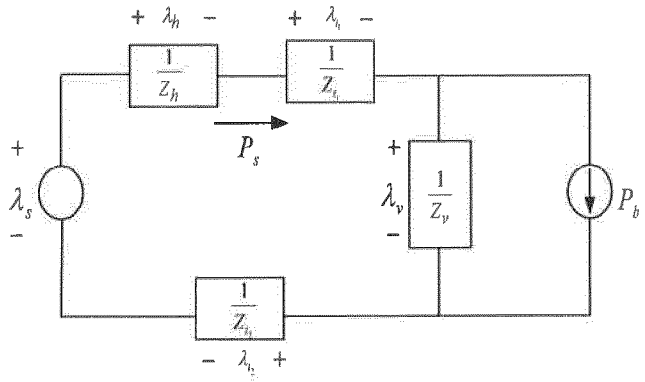
+ λ_h −
+ λ_t −
$\frac{1}{Z_h}$
$\frac{1}{Z_t}$
P_s
+
λ_s
−
+
λ_v
−
$\frac{1}{Z_v}$
P_b
$\frac{1}{Z_b}$
− λ_b +

FIG.2c

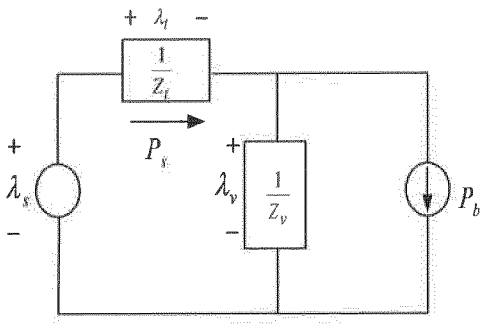
+ λ_t −
$\frac{1}{Z_t}$
P_s
+
λ_s
−
+
λ_v
−
$\frac{1}{Z_v}$
P_b

FIG.2d

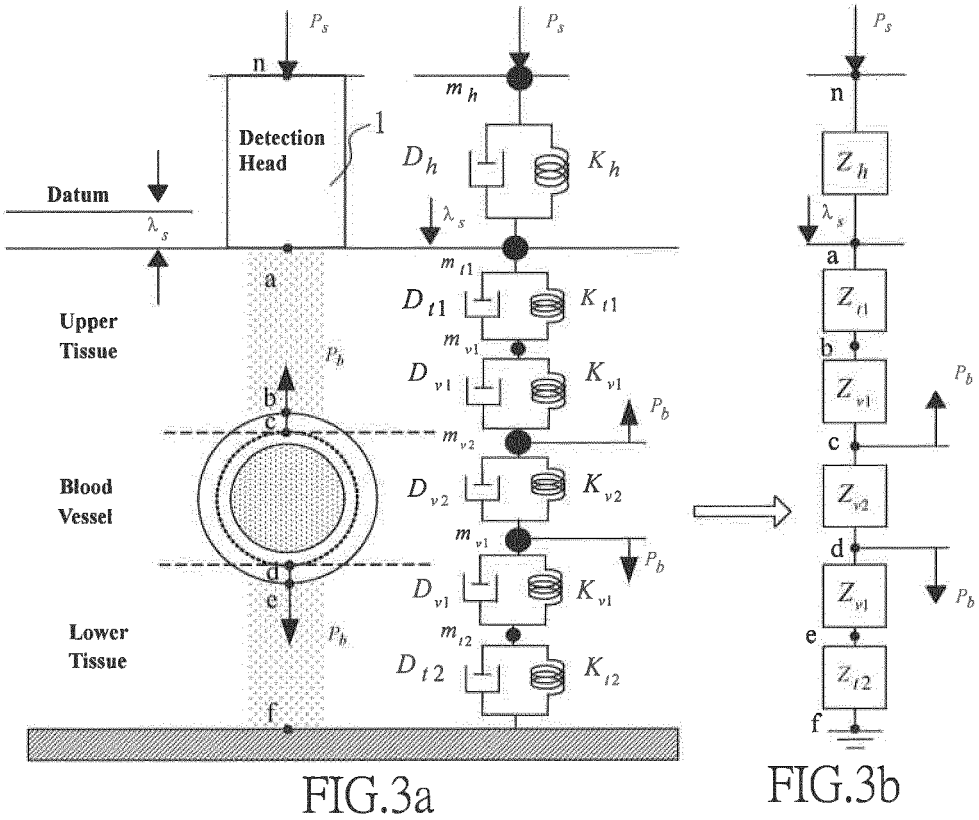
Detection Head
1
Datum
Upper Tissue
Blood Vessel
Lower Tissue
n
a
b
c
d
e
f
P_s
P_b
λ_s
m_h
D_h
K_h
m_{t1}
D_{t1}
K_{t1}
m_{v1}
D_{v1}
K_{v1}
m_{v2}
D_{v2}
K_{v2}
m_{t2}
D_{t2}
K_{t2}
Z_h
Z_{t1}
Z_{v1}
Z_{v2}
Z_{t2}

FIG.3a

FIG.3b

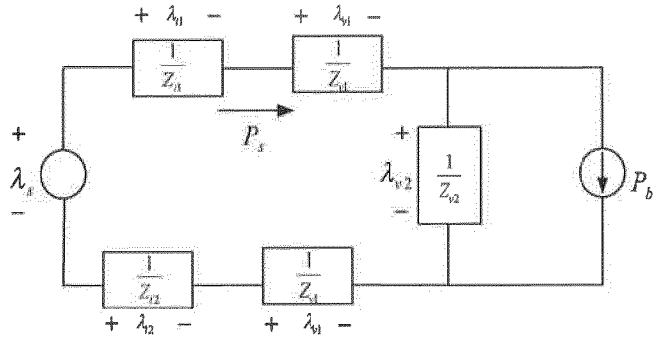

FIG.3c

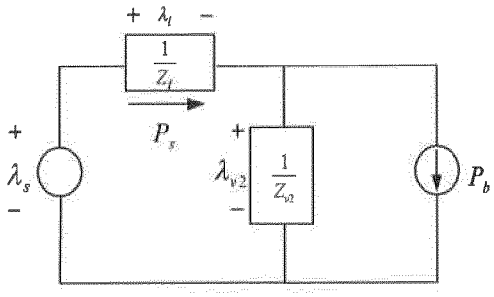

FIG.3d

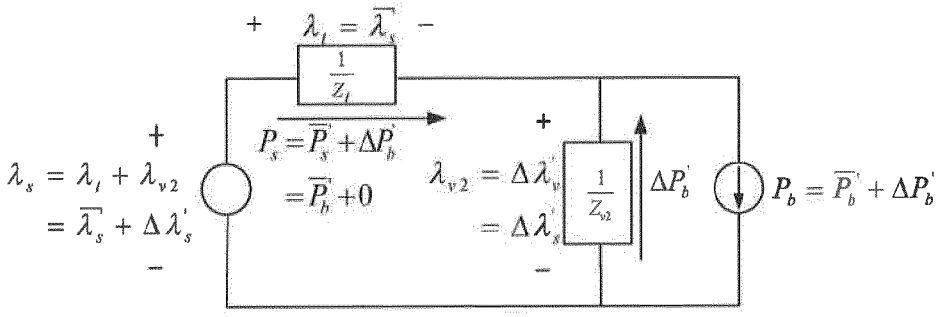

FIG.3e

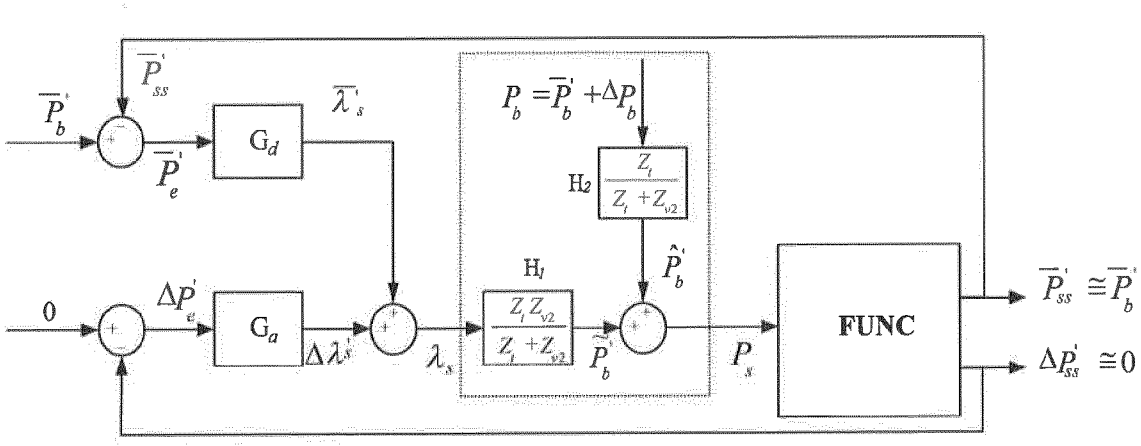
$\overline{P}_{ss}^{'}$
$\overline{P}_{b}^{'}$
$\overline{P}_{e}^{'}$
G_d
$\overline{\lambda}_{s}^{'}$
$P_b = \overline{P}_b^{'} + \Delta P_b$
H_2
$\frac{Z_l}{Z_l + Z_{v2}}$
$\hat{P}_b^{'}$
H_1
0
$\Delta P_e^{'}$
G_a
$\Delta \lambda_s^{'}$
λ_s
$\frac{Z_l Z_{v2}}{Z_l + Z_{v2}}$
$\tilde{P}_b^{'}$
P_s
FUNC
$\overline{P}_{ss}^{'} \cong \overline{P}_b^{'}$
$\Delta P_{ss}^{'} \cong 0$

FIG.4

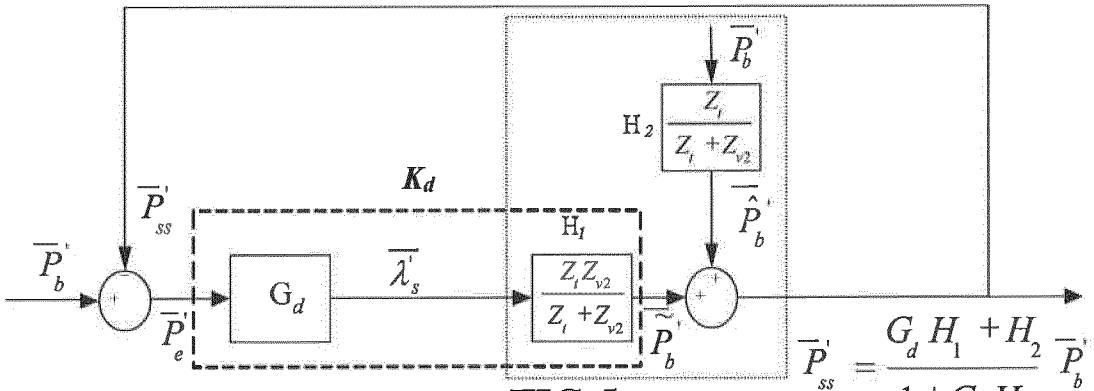
$\overline{P}_b'$
$H_2 \frac{Z_t}{Z_t + Z_{v2}}$
K_d
$\overline{P}_{ss}'$
H_1
$\overline{\hat{P}}_b'$
$\overline{P}_b'$
G_d
$\overline{\lambda}_s'$
$\frac{Z_t Z_{v2}}{Z_t + Z_{v2}}$
$\overline{P}_e'$
$\overline{\tilde{P}}_b'$
$\overline{P}_{ss}' = \frac{G_d H_1 + H_2}{1 + G_d H_1} \overline{P}_b'$

FIG.5

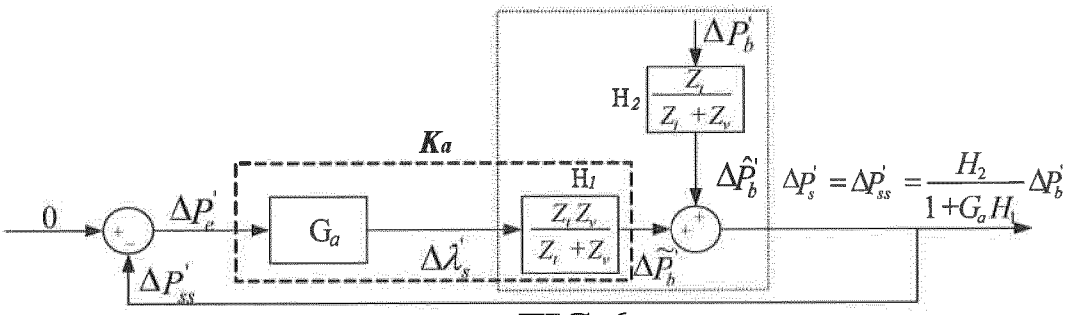
$\Delta P_b'$
$H_2 \frac{Z_t}{Z_t + Z_v}$
K_a
H_1
$\Delta \hat{P}_b'$
$\Delta P_s' = \Delta P_{ss}' = \frac{H_2}{1 + G_a H_1} \Delta P_b'$
0
$\Delta P_e'$
G_a
$\frac{Z_t Z_v}{Z_t + Z_v}$
$\Delta \lambda_s'$
$\Delta \tilde{P}_b'$
$\Delta P_{ss}'$

FIG.6

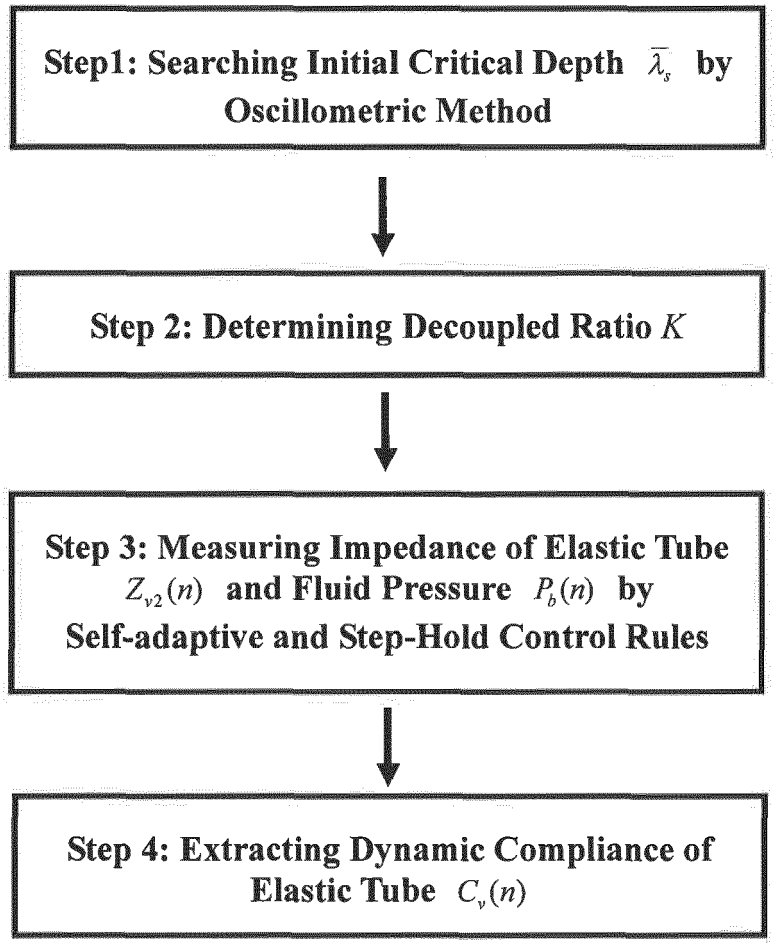
Step1: Searching Initial Critical Depth $\overline{\lambda}_s$ by Oscillometric Method
Step 2: Determining Decoupled Ratio K
Step 3: Measuring Impedance of Elastic Tube $Z_{v2}(n)$ and Fluid Pressure $P_b(n)$ by Self-adaptive and Step-Hold Control Rules
Step 4: Extracting Dynamic Compliance of Elastic Tube $C_v(n)$

FIG. 7a

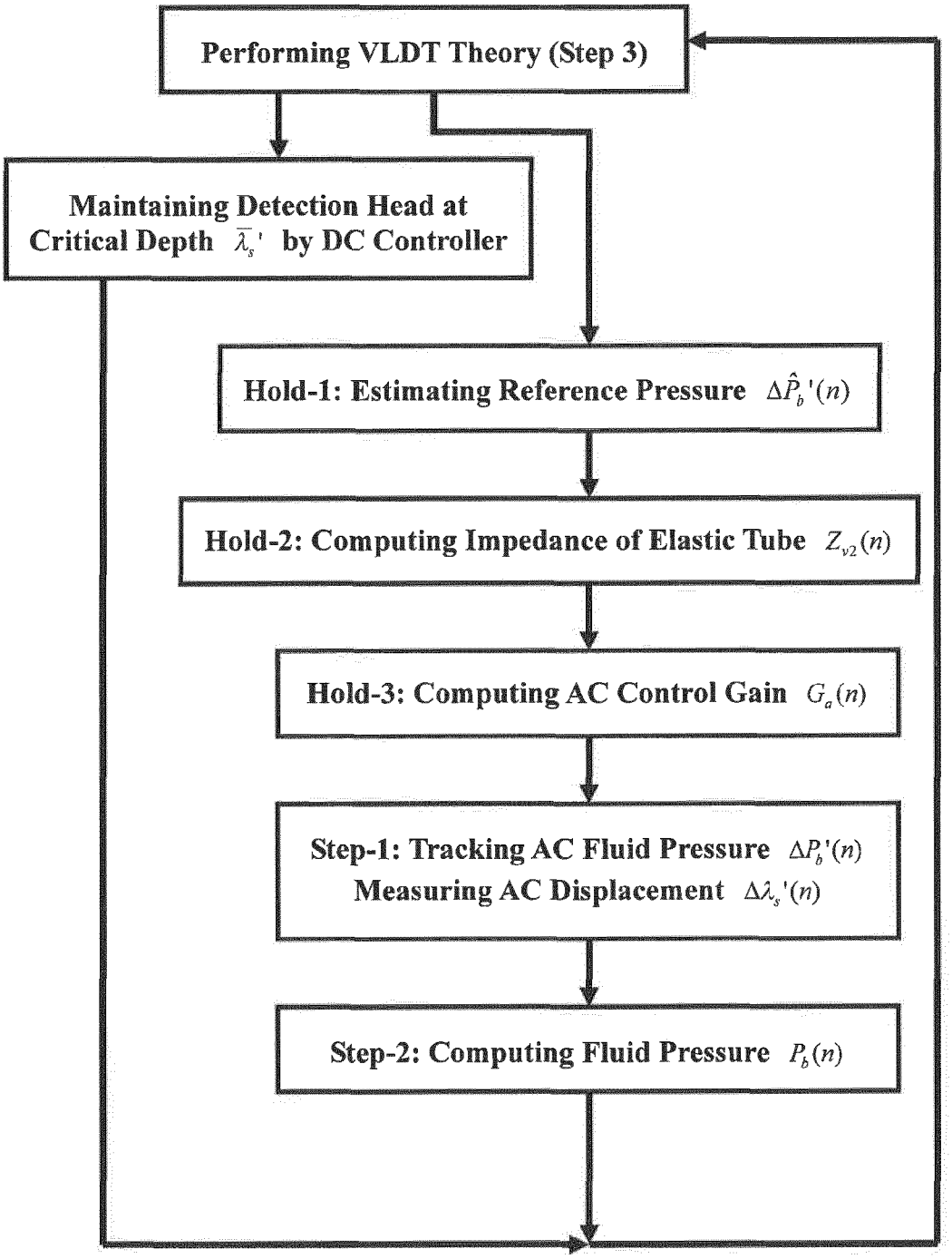
Performing VLDT Theory (Step 3)
Maintaining Detection Head at Critical Depth $\overline{\lambda}_s'$ by DC Controller
Hold-1: Estimating Reference Pressure $\Delta\hat{P}_b'(n)$
Hold-2: Computing Impedance of Elastic Tube $Z_{v2}(n)$
Hold-3: Computing AC Control Gain $G_a(n)$
Step-1: Tracking AC Fluid Pressure $\Delta P_b'(n)$
Measuring AC Displacement $\Delta\lambda_s'(n)$
Step-2: Computing Fluid Pressure $P_b(n)$

FIG. 7b

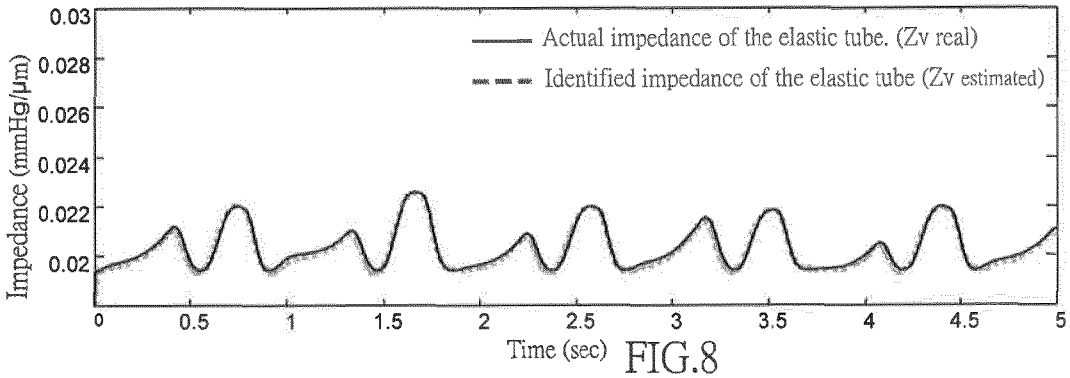
Actual impedance of the elastic tube. (Zv real)
Identified impedance of the elastic tube (Zv estimated)
Impedance (mmHg/μm)
0.03
0.028
0.026
0.024
0.022
0.02
0
0.5
1
1.5
2
2.5
3
3.5
4
4.5
5
Time (sec)

FIG.8

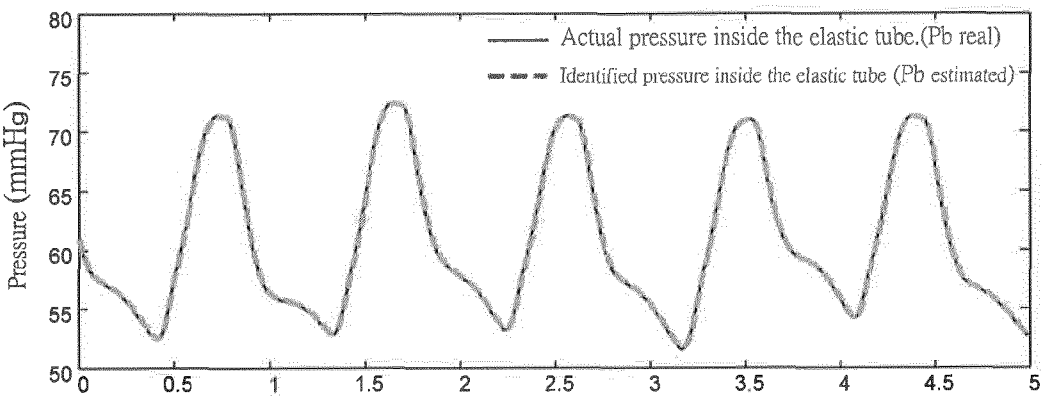
Actual pressure inside the elastic tube.(Pb real)
Identified pressure inside the elastic tube (Pb estimated)
Pressure (mmHg)
80
75
70
65
60
55
50
0
0.5
1
1.5
2
2.5
3
3.5
4
4.5
5
Time (sec)

FIG.9

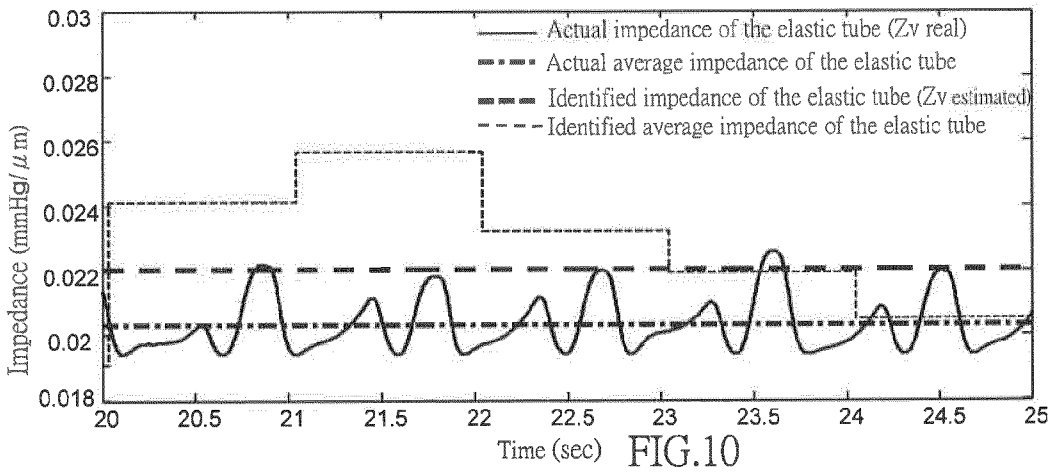
Actual impedance of the elastic tube (Zv real)
Actual average impedance of the elastic tube
Identified impedance of the elastic tube (Zv estimated)
Identified average impedance of the elastic tube
Impedance (mmHg/ μ m)
0.03
0.028
0.026
0.024
0.022
0.02
0.018
20
20.5
21
21.5
22
22.5
23
23.5
24
24.5
25
Time (sec)

FIG.10

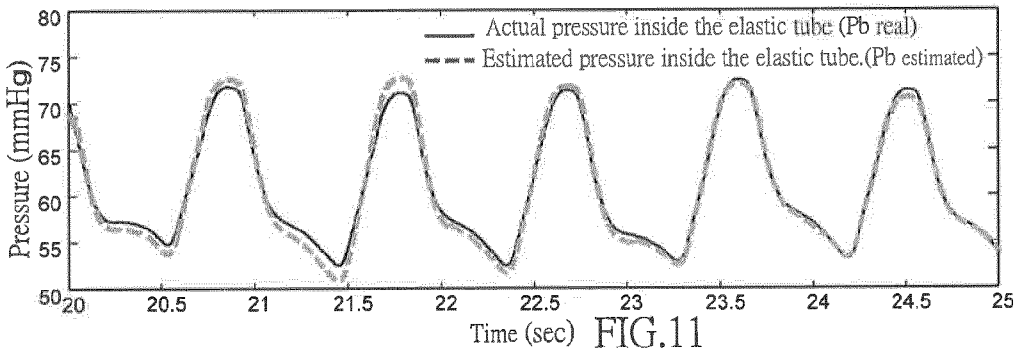
Actual pressure inside the elastic tube (Pb real)
Estimated pressure inside the elastic tube.(Pb estimated)
Pressure (mmHg)
80
75
70
65
60
55
50
20
20.5
21
21.5
22
22.5
23
23.5
24
24.5
25
Time (sec)

FIG.11

Europäisches Patentamt
European Patent Office
Office européen des brevets

EUROPEAN SEARCH REPORT

Application Number
EP 13 15 9546

DOCUMENTS CONSIDERED TO BE RELEVANT

Category	Citation of document with indication, where appropriate, of relevant passages	Relevant to claim	CLASSIFICATION OF THE APPLICATION (IPC)
A	US 2009/069699 A1 (LIN ALBERT CHIN-YUH [TW]) 12 March 2009 (2009-03-12) * the whole document * -----	1-4	INV. G01L9/00 A61B5/022 A61B5/00

TECHNICAL FIELDS SEARCHED (IPC)

A61B
G01L

The present search report has been drawn up for all claims

Place of search	Date of completion of the search	Examiner
The Hague	10 September 2013	Reto, Davide

CATEGORY OF CITED DOCUMENTS

X : particularly relevant if taken alone
Y : particularly relevant if combined with another document of the same category
A : technological background
O : non-written disclosure
P : intermediate document

T : theory or principle underlying the invention
E : earlier patent document, but published on, or after the filing date
D : document cited in the application
L : document cited for other reasons

& : member of the same patent family, corresponding document

EPO FORM 1503 03.82 (P04C01)

ANNEX TO THE EUROPEAN SEARCH REPORT ON EUROPEAN PATENT APPLICATION NO. EP 13 15 9546

This annex lists the patent family members relating to the patent documents cited in the above-mentioned European search report. The members are as contained in the European Patent Office EDP file on
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10-09-2013

Patent document cited in search report	Publication date	Patent family member(s)	Publication date
US 2009069699 A1	12-03-2009	JP 2009066384 A	02-04-2009
		TW 200911199 A	16-03-2009
		US 2009069699 A1	12-03-2009

EPO FORM P0459

For more details about this annex : see Official Journal of the European Patent Office, No. 12/82

专利名称(译)	一种测量弹性管内流体实时连续压力和弹性管动态顺应性的无创方法和装置		
公开(公告)号	EP2778643A1	公开(公告)日	2014-09-17
申请号	EP2013159546	申请日	2013-03-15
[标]申请(专利权)人(译)	龚BU设计		
申请(专利权)人(译)	龚BU设计公司		
当前申请(专利权)人(译)	龚BU设计公司		
发明人	LIN,, ALBERT CHIN-YUH		
IPC分类号	G01L9/00 A61B5/022 A61B5/00		
CPC分类号	A61B5/022 A61B5/6843 G01L9/0026		
外部链接	Espacenet		

摘要(译)

本发明提供了一种无创的方法和装置，用于测量弹性管中波动的液体的实时连续压力和弹性管的动态顺应性，其中使用了VLDT（血管加载解耦技术）的理论。在搜索初始临界深度并确定解耦比率之后，DC控制器产生DC控制增益以将弹性管保持在临界深度，并且AC控制器采用自适应和步进保持控制规则来产生弹性脉动。管不受周围组织的影响，能够测量实时连续流体压力和弹性管的动态顺应性。

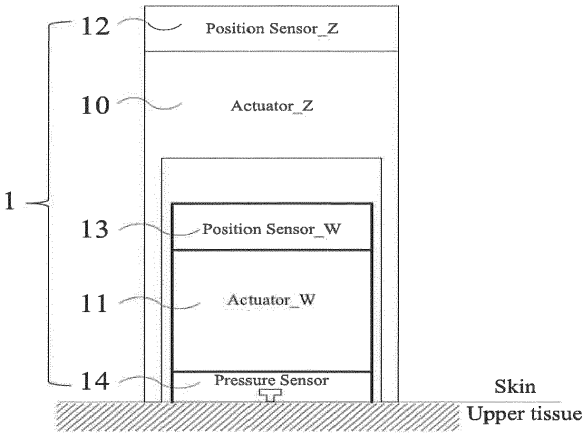

FIG.1