DESIGN AND FABRICATION OF FLOW PHANTOM FOR PULSED ARTERIAL SPIN LABELING AT 3 T MAGNETIC RESONANCE IMAGING

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By

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LIST OF ABBREVIATIONS

3D	Three dimensional
ASL	Arterial spin labeling
bSSFP	Balanced steady-state free precession
CASL	Continuous arterial spin labeling
CBF	Cerebral blood flow
CEMRA	Care contrast-enhanced MRA
DIPLOMA	Double inversions with proximal labeling
EPI	Echo planar imaging
EPISTAR	Echo planar imaging and signal targeting with alternating RF
FAIR	Flow-sensitive alternating inversion recovery
FLASH	Fast low angle shot
FOV	Field of view
MRA	Magnetic resonance angiography
MRI	Magnetic resonance imaging
MT	Magnetization transfer
PASL	Pulsed arterial spin labeling
pCASL	Pseudocontinuous arterial spin labeling
PETRA	Pointwise encoding time reduction with radial acquisition
PICORE	Proximal inversion with a control for off-resonance effects
PULSAR	Pulsed star labeling of arterial regions
RF	Radiofrequency
ROI	Region of Interest
SE	Spin echo
SENSE	Sensitivity encoding
SNR	Signal-to-noise ratio
TE	Echo time
TOF	Time-of-flight
TR	Repetition time

DESIGN AND FABRICATION OF FLOW PHANTOM FOR PULSED ARTERIAL SPIN LABELING AT 3 T MAGNETIC RESONANCE IMAGING

ABSTRACT

Pulsed arterial spin labeling (PASL) is a main type of ASL technique that use magnetically labeled arterial blood water as tracer to produce images. The aim of this study was to design and fabricate a flow phantom that can be used as a component of PASL study to evaluate optimal Magnetic Resonance (MR) imaging parameters that can be used to obtain optimal MR image quality. In this study, a Perspex based flow phantom was designed with a set of tubes that mimicking carotid arteries in both adults and pediatric; and presented with 50% and 75% stenoses. A mixture of 60:40 distilled water and glycerol was used to mimic blood. The image acquisition of the phantom was performed by using 3 T MR. The phantom used 16-channel head and neck coil and scanned using PASL technique in combination with ASL multiphase and single shot echo planar imaging (EPI) sequence and sensitivity encoding (SENSE). The field of view (FOV) and slice thickness increase the signal-to-noise ratio (SNR) by increasing the voxel size. The MR images quality was evaluated by measuring the SNR. Study found that the highest SNR by average was obtained by using imaging parameters: FOV: 320 x 320 mm^2 and slice thickness: 9 mm. In conclusion, optimal image quality depends upon best scanning parameter choices. Therefore, this study may be served as a guideline for the specification of ASL application in future.

REKAAN DAN FABRIKASI FANTOM BERALIRAN UNTUK TEKNIK PELABELAN ARTERI BERPUTAR SECARA DENYUT DALAM 3 T IMBASAN MAGNETIK BERESONANSI

ABSTRAK

Pelabelan arteri berputar secara bernadi (PASL) merupakan sejenis teknik ASL yang menggunakan cecair darah arteri yang telah dilabelkan secara magnetik sebagai alat pengesanan untuk menghasilkan imej. Kajian ini bertujuan untuk mereka dan memfabrikasi sebuah fantom beraliran yang boleh digunakan sebagai komponen dalam penyelidikan PASL untuk mengenalpasti parameter terbaik yang boleh digunakan untuk menghasilkan imej berkualiti tinggi. Di dalam penyelidikan ini, sebuah fantom beraliran Perpex telah direka dengan satu set tiub yang menyerupai arteri karotid orang dewasa dan kanak-kanak; dan juga mempunyai 50% dan 75% bahagian berstenosis. Satu campuran cecair daripada 60:40 air suling dan gliserol telah digunakan untuk menyerupai darah. Pemerolehan imej fantom dilakukan dengan menggunakan 3 T imbasan magnetik beresonansi (MRI). Fantom menggunakan gegelung kepala dan leher bersaluran 16 dan diimbas mengunakan teknik PASL dengan gabungan multifasa ASL dan urutan EPI dan SENSE. Medan penglihatan (FOV) dan ketebalan lapisan dapat meningkatkan nisbah isyarat kepada bunyi (SNR) dengan meninggikan saiz voxel. Kualiti imej telah dinilai dengan nilai SNR. Hasil kajian mendapati nilai SNR tertinggi secara purata telah diperolehi dengan menggunakan parameter pengimejan: FOV: 320 x 320 mm^2 dan ketebalan lapisan: 9 mm. Secara konklusinya, penghasilan imej berkualiti optimal bergantung pada pemilihan imbasan parameter terbaik. Oleh itu, kajian ini mungkin dapat dijadikan garis panduan untuk spesifikasi aplikasi ASL di masa hadapan.

ACADEMIC CONTRIBUTION

These are the lists of papers that have been submitted to journals for publication:

1. **Awatif M. R.**, Norain Y., Jihan M. Z., Fara A., and Zainon R., Design and Fabrication of Flow Phantom for Pulsed Arterial Spin Labeling at 3 T Magnetic Resonance Imaging, (2015). Paper was submitted to Journal of Instrumentation.

2. Jihan M. Z., Fara A., **Awatif M. R.**, Norain Y., and Zainon R., Design and Fabrication of Arterial Spin Labeling Magnetic Resonance Imaging Flow Phantom for Evaluation of Imaging Parameters, (2015). Paper was submitted to Journal of Instrumentation.

3. Norain Y., **Awatif M. R.**, Jihan M. Z., Fara A., and Zainon R., Evaluation of pseudo-Continuous Arterial Spin Labeling (pCASL) in 1.5 T Magnetic Resonance Imaging: Phantom Study, (2015). Paper was submitted to MRI Journal.

CHAPTER 1

INTRODUCTION

1.1 Background of the study

As a type of magnetic resonance (MR) imaging technique, arterial spin labeling (ASL) was introduced over 20 years ago (Alsop *et al.*, 2014; Pollock *et al.*, 2009). The technique potents to evaluate hemodynamic perfusion and may be useful in explaining the relationship between the vasculature, perfusion and brain function (Van Laar *et al.*, 2008). The technique ables to measure changes of cerebral blood flow (CBF), thus become one of capable technique used for diagnosing cerebrovascular disease like carotid artery stenosis and for the evaluation of pre- and post- treatment. In contrast to the conventional MR imaging, the ASL is technically non invasive. Instead of using exogenous radioactively labeled water or exogenous paramagnetic contrast agent like gadolinium, the technique use endogenous magnetically labeled arterial blood water as the tracer (Alsop *et al.*, 2014; Martirosian *et al.*, 2010). This main advantage of ASL enable acquiring perfusion data in pediatric patient and adult patient contraindicated to the radioactive agent especially for patient having renal problem (Sadowski *et al.*, 2007).

In this technique, the arterial blood water proton will be magnetically labeled just below the region of interest. Then, an inversion of net magnetisation of blood water will be resulted from the application of radiofrequency (RF) inversion pulse. After a period of time, called the transit time, the magnetically arterial blood water will exchanges with the tissue water at the region of interest resulting in total tissue magnetisation alteration and reduction which then provide the signal and image intensity, where the first image, called label or tag image obtained. To create another image, called control image, the experiment will be repeated without arterial blood label. Then, the perfusion image will be obtained from the subtraction of these two images. (Petcharunpaisan *et al.*, 2010; Wintermark *et al.*, 2005). Usually, the image acquisition is between 5 to 10 minutes by depending on the scanner quality, magnetic field (defined as Tesla (T)), RF coil sensitivity and the application of sensitivity encoding (SENSE) for fast MRI (Wintermark *et al.*, 2005).

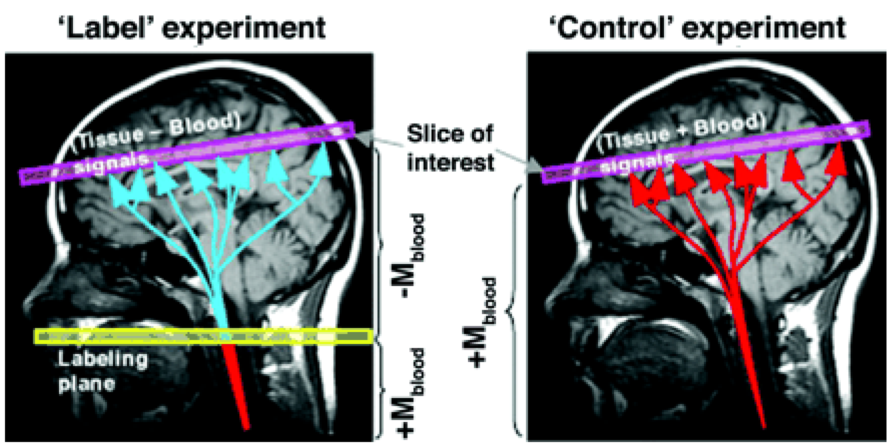


Figure 1.1 The main principle of ASL technique, where the subtraction of first image (on the left) called tag or label image with second image (on the right) called control image yield perfusion information, the quantitative CBF. Retrieved from Wintermark *et al.* (2005).

The ASL techniques are mainly branched into two types, which are pulsed arterial spin labelling (PASL) and continuous arterial spin labelling (CASL) (Essig *et al.*, 2013). These two techniques are basically differ in how the RF pulse applied during labeling and their tagging based. In PASL, short RF pulse is selectively applied to just specific regions and the tagging based on the location. Meanwhile, the RF pulse is applied in a continuous manner in CASL and the tagging based on location and velocity. (Petcharunpaisan *et al.*, 2010). The PASL technique offer high inversion efficiency and less RF power use. As the tag can be applied more closely in the technique, the CASL produces higher overall tagging efficiency and signal-to-noise ratio (SNR). Based on these two techniques, the current technique known as pseudocontinuous ASL was developed. This technique implies the short RF pulse version in PASL and producing a high signal to noise ratio (SNR) as in CASL (Petcharunpaisan *et al.*, 2010).

Pulsed Arterial Spin Labeling

PASL tagging based on the location only without involving any velocity. Basically, the technique is very efficient in implies the inversion of region of interest by using the RF pulse. According to Alsop *et al.*, (2014), there are several conditions to be applied for the implementation of the technique. Firstly, to get an efficient inversion (> 95%), the use of adiabatic inversion pulses is recommended as such inversion allows the RF pulses be insensitive to inhomogeneities. Secondly, it is must to ensure that the total RF applied during tagging and control image equal in order to minimise magnetisation transfer effect. Moreover, a proper selection of slabs thickness at optimal size must be made to avoid overlapping with imaging volume. The slab thickness should be as large as possible, the recommended thickness is between 15 to 20 cm. There are various labeling methods applicable to be used in PASL including flow-sensitive alternating inversion recovery (FAIR), echo planar imaging and signal targeting with alternating RF (EPISTAR), proximal inversion with a control for off-resonance effects (PICORE), pulsed star labeling of arterial regions (PULSAR) and double inversions with proximal labeling of both tag and control images (DIPLOMA). These labeling methods have potential differences in signal from inflowing distal spin, thus the applicants must choose the method properly in order to obtain ASL signal. One of the disadvantages of PASL is it creates labeling bolus with an unknown and short temporal width; make it harder to be used for the quantification of CBF with a single delay time (Alsop *et al.*, 2014).

1.2 Problem statement

As the ASL is technically non-invasive and can perform similar function as in other imaging technique, demonstrated by abundance of past and recent studies, this technique should be a standard practice in wide clinical and research practices. However, the situation is otherwise as the technique presented with wide range of choices and become harder situation for the clinicians and new researcher to choose appropriate method to be applied clinically and in research practices (Alsop *et al*., 2014). In order to contribute more knowledge regarding ASL technique, this study was conducted by using PASL technique with the aims to obtain useful information for better understanding of ASL and specification of ASL application in future.

Evaluation of optimal MR imaging parameters are usually time consuming and require repetitive acquisition of data. Such process is inconvenience to involve living human subject. Therefore, phantom study is the best alternative measure as it more readily available. According to Summers *et al.,* (2005), an ideal phantom should be easily assembled, portable, workable, testable and applicable for multi trials. In this study, a flow phantom was designed and fabricated as a component of PASL study to test the imaging parameters. As the world's population is in transition of age structure towards aging population, defined by the expectation of the people who are over 60 years old which are 11.7% in 2013 to be grow to 21.1% by 2050 and as the number of the older persons are projected to exceed the number of children (United Nations, 2013); along with the availability of non-invasive imaging studies, the cerebrovascular disease like carotid artery stenosis which is mainly associates with stroke, is a disease commonly seen in medical or clinical settings. As a part of multi trials purpose, the phantom was designed with a set of tubes that mimicking carotid arteries in both adults and pediatric; and presented with 50% and 75% stenotic parts which are incorporated for use in future studies of the disease.

1.3 Objectives

1.3.1 General Objective:

To design and fabricate a flow phantom for PASL at 3 T Magnetic Resonance imaging

1.3.2 Specific Objectives:

- To design and fabricate a flow phantom for PASL MRI
- To evaluate the optimal imaging parameters for PASL to obtain optimal image quality by using the fabricated phantom

1.4 Hypothesis

The designated and fabricated flow phantom can be used to evaluate imaging parameter in PASL study. The optimal imaging parameters for optimisation of MR image quality can be obtained by using fabricated phantom scanned with PASL technique at 3 T MRI.

CHAPTER 2

LITERATURE REVIEW

2.1 Pros, cons and clinical applications of ASL

From pros of view, the ASL technique can provide absolute quantification. This statement was proved widely by past literatures. According to Wintermark *et al.*, (2005), it showed that the estimation of blood flow in gray matter by using ASL technique was correct when computer simulations by using an extensive model of ASL technique have been studied by St Lawrence *et al.*, (2000). This finding also was supported by the result of comparison study between ASL technique with other non-nuclear MR imaging technique addressed by Buxton *et al.*, (1998) and Ye *et al.*, (2000), According to Parkes *et al.*, (2004), the quantification resulted from ASL scanning shows only less than 10% changes when repeated scanning implemented on the same subject. Furthermore, acquiring perfusion data can be performed on any patient without contraindicated to MR imaging; and in pediatric or infant as it requires non-invasive procedure and lack from ionising radiation exposure (Sadowski *et al.*, 2007; Wintermark *et al.*, 2005).

From cons of view, Wintermark *et al.*, (2005) stated that the signal difference between tag and control image of ASL is low (≈1%), thus the technique requires a very high signal-to-noise ratio in order to obtain perfusion data. Besides that, the technique is very sensitive to subject movement due to the subtraction manner and can contribute for visibility of motion artifacts in the images obtained. Because of the MR imaging sequence suppress the signal from large vessels, ASL was believed not sensitive to large arteries or veins as the signal only obtained from water found in small vessels and surrounding tissue. The ASL provides low signal-to- noise ratio (SNR) which can defect the image quality of perfusion (Alsop *et al.*, 2014; Borogovac, 2012; Pollock *et al.*, 2009). This limitation can be solved by increasing the strength of magnetic field used and changing the imaging parameters (Pollock *et al*., 2009, Wang *et al*.., 2002). For image resolution, the ASL yields image perfusion with higher spatial and temporal resolution than any other current technique. (Borogovac, 2012; Vita *et al*., 2003). Furthermore, Vita *et al.*, (2003) also found that ASL become a more sensitive method at higher magnetic field as the result from increasing spin-lattice relaxation time for blood perform cerebral blood flow measurement. It was believed that ASL technique reached the level of high usefulness that suit to be applied in clinical and research, as it already passed the 20 years aged. However, the situation is otherwise as the technique presented with wide range of choices and become harder situation for the clinicians and new researcher to choose appropriate method, thus result in low availability in clinical and research practices (Alsop *et al*., 2014).

Although ASL technique has low availability in widespread clinical practices, the technique has been demonstrated for several acute and chronic cerebrovascular diseases like stroke and transient ischemic attack; temporal lobe epilepsy and brain tumor perfusion (Wolf *et al.,* 2003). The quantifications facilitate in recognition of state like hypercapnia or diffuse hypoxic/anoxic injury; and also in comparison between pre and post treatment like after patients administered by cerebrovascular dilator such as acetazolamide (Wintermark *et al.*, 2005); or during pre and post neurointerventional procedure such as carotid endarterectomy and stenting (Ances *et al.*, 2004); thrombolysis, migraine or seizure therapy for regional assessments (Pollock *et al.*, 2009). In comparison with conventional MR imaging that using radioactive contrast, ASL perfusion imaging has equally effective in determining tumor grade in a study of malignant gliomas (Warmuth *et al.*, 2003). According to Chalela *et al.*, (2000), CBF reduction in symptomatic hemisphere that has been measured by continuous ASL is well correlated to National Institutes of Health Stroke Score.

2.2 Measures to improve image acquisition in ASL technique

As the ASL technique only use magnetically labeled blood water as tracer, a drawback of ASL technique can be seen through low SNR of image acquisition. According to Pollock *et al.*, (2009), Wang *et al.*, (2002) and Wolf and Detre (2007), higher magnetic field is one of the measures that can be used to increase image SNR. Higher magnetic field strength, defined as Tesla does not only increasing SNR, but also can causes T1 lengthens and allowing the accumulation of more spin label that catch the signal, thus resulting in higher signal achieved (Wang *et al.*, 2002). It is recommended to use 3 T and above of MR imaging magnetic field for better image acquisition in ASL (Wolf and Detre, 2007).

The second measure that can be taken to increase SNR is by using a phased array receiver. Although phased array coil produces inhomogeneities signal at particular spaces, the absolute CBF quantification still can be obtained as several calibration processes involved during the quantification (Wang *et al*., 2005). When the phased array optimised for parallel imaging to reduce scan time, there will be reduction of SNR. However, the

SNR will be increased again as the shortened TE applied along with the reduction of distortion caused by susceptibility artifact (Wang *et al*., 2005). The use of the combination of high magnetic field and array coils can contribute to the unnecessary sources of the noise in ASL. Therefore, the use of background suppression that has been popular nowadays is the best way to improve ASL robustness and provide true SNR (Wolf and Detre, 2007).

According to Wolf and Detre (2007), in order to compensate the low SNR in ASL, an imaging sequence that has high SNR will be employed together along the technique implementation. The commonly used echoplanar imaging (EPI) sequence has high SNR and can reduce the potential motion artifacts. However, EPI can produce another type of artifacts, called as susceptibility artifacts. Recently, fast 3D sequences have been introduced to be implemented along with ASL technique. 3D sequences can improve SNR and can reduce the distortion visible from susceptibility artifacts. Another advantage of the new implemented sequence is it facilitates the use of background suppression to reduce the static brain signal (Ye *et al.*, 2000). Background suppression along with appropriately timed inversion pulses can increase the ASL effect and increasing the sensitivity for CBF quantification (Wolf and Detre, 2007).

2.3 ASL phantom study

In 2003, there was a study conducted to demonstrate improved compensation of magnetisation transfer (MT) in new PASL technique for multislice perfusion-weighted imaging by using phantom. In this study, a 3% agarose phantom with tap water doped by 0.45 mM/L of copper sulfate to maintain T_1 = 1056.0 ± 5.5 ms was used to compare the abilities of DIPLOMA, PICORE, and EPISTAR to avoid MT effects for single-slice acquisition. Study found that the improvement of MT with the new method increasing image contrast, mean signal intensity and provide better signal uniformity across the slices, thus improved the accuracy of brain perfusion quantification (Jahng *et al.,* 2003).

In 2007, there was a pulsed ASL study conducted by Noguchi *et al.* on a flow phantom. Their aim was to confirm the validity of quantitative perfusion imaging resulted from the used of Q2TIPS sequence of the PASL that have been introduced by Luh *et al.* (1999). The Q2TIPS that has a saturation pulse after labeling pulse was produced by the modification of QUIPPS and QUIPPS II. In order to make the confirmation, they compare the true and theoretical flow rates in a flow phantom. The flow phantom was designed by using 40 mm diameter plastic syringe filled with plastic beads and plastic tubes 4 mm diameter. By using adjusted flow rate of constant flow pump between 0 cm/s and 2.61 cm/s, 8 litre Gd-DTPA-doped water solution (0.1 Mm) was circulated from tank to syringe through a plastic tube. The phantom was scanned by using Q2TIPS sequence parameter, TI_1=50 ms and TI_2=1400 ms. As the result, they found a good linear relationship between theoretical and true flow rate, which means the quantitative perfusion imaging by using Q2TIPS sequence of PASL accurate and the validity was confirmed (Noguchi *et al.*, 2007).

In a recent study did by Koktzoglou *et al.*, (2015), they used eight vascular models of carotid artery stenosis to quantify the accuracy of a new implemented pseudocontinuous ASL, the three-dimensional (3D) radial ASL magnetic resonance angiography (MRA). In the study, the models were scanned with 1.5 T MRA using flow rates of 100-400 mL/min as found in internal carotid artery. The quantification of 3D ASL imaging using fast low angle shot (FLASH), balanced steady-state free precession (bSSFP) and pointwise encoding time reduction with radial acquisition (PETRA) were compared with standard-of-care contrast-enhanced MRA (CEMRA) and Cartesian time-of-flight (TOF) MRA protocols. From the comparison, they found that the ASL MRA, especially the ASL FLASH and ASL PETRA can improve the hemodynamic quantification in carotid artery stenosis (50% and 70%) compared to TOF MRA (Koktzoglou *et al.*, 2015).

CHAPTER 3

METHODOLOGY

3.1 Research Design

The study was conducted based on experimental study design in Imaging Unit of Hospital Universiti Sains Malaysia, Kubang Kerian, Kelantan.

3.2 Design and fabrication of phantom

An ideal phantom should be easily assembled, portable, workable and testable (Summers *et al.,* 2005). In order to fulfill these criterions and to produce an applicable object for multi trials, a flow phantom was designed to mimic cerebral blood vessels, the carotid artery of adult and pediatric; presented with stenoses and without stenoses region. The design of the phantom was based on Summers *et al.,* (2005). However, the current phantom was made up of Perspex, not the silicone polymer as proposed by the past researchers. The Perspex was selected as the phantom's material because it has low mass magnetic susceptibility of 0.50 m^2 / kg. Moreover, the Perspex is a stable, safe and easy to handle material.

Initially, the phantom was designed with dimension 240 x 240 x 25 mm^3 size (Refer Figure 3.1). It is comprised of a set of straight tubes and a set of constant diameter tubes in the shape of U bend. One of the straight tubes is 5 mm diameter while the other one straight tube and another two tubes with U bend are 8 mm diameter. The 5 mm diameter tube represents pediatric carotid artery while 8 mm diameter tubes represents adult carotid artery. The two tubes with U bend have 50% and 75% of stenosis by diameter respectively.

In order to fit the head and neck coil of the major MR scanner manufacturers, the initial design was modified by dividing the phantom into two parts (Figure 3.2 and Figure 3.3). Each of the parts has dimension of 240 x 120 x 25 mm^3 size respectively.

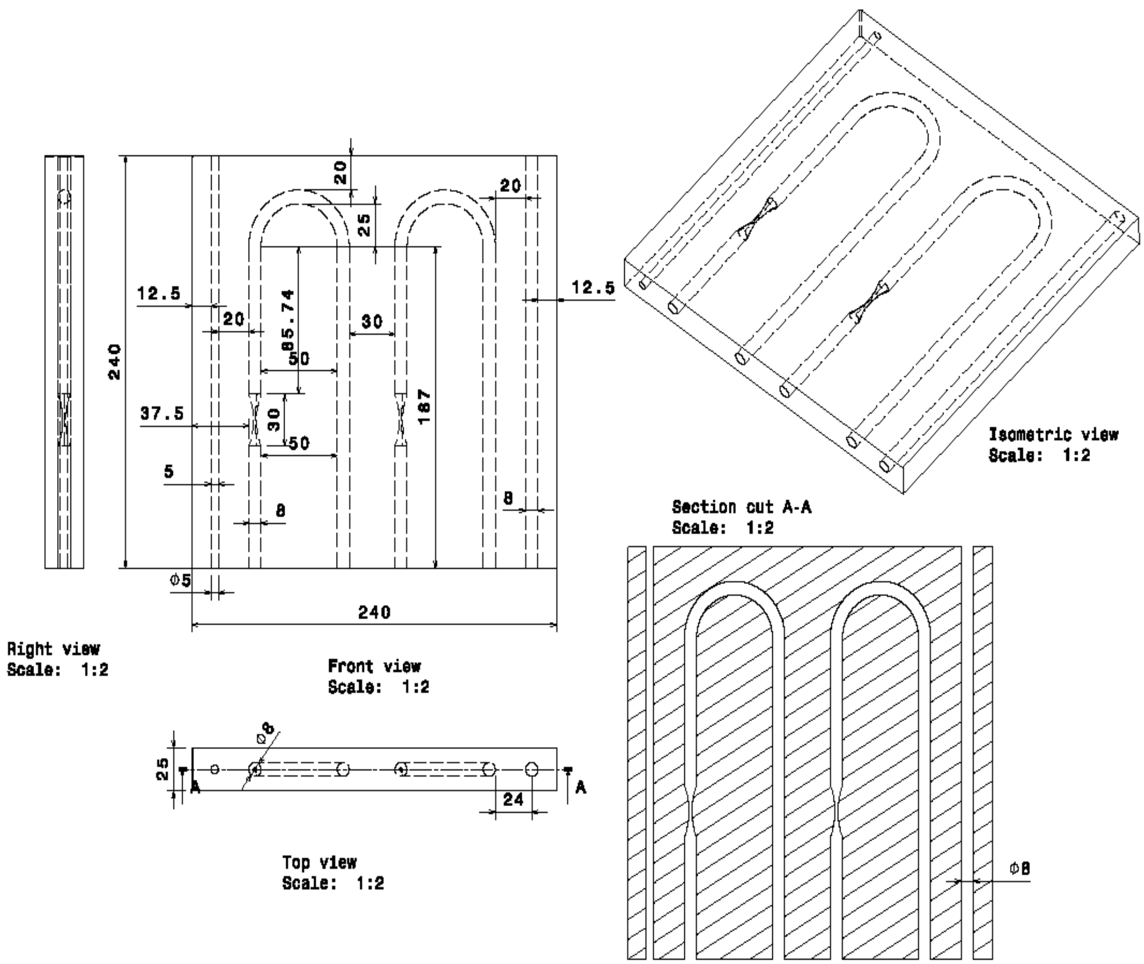
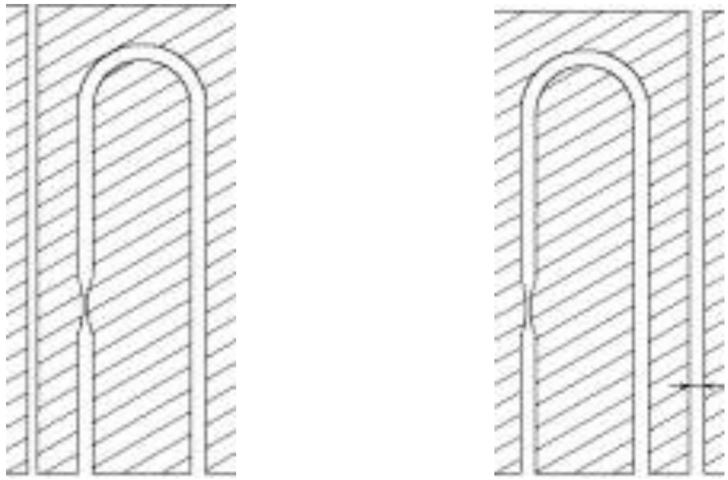


Figure 3.1 Schematic diagram of the initial design of the phantom



(a) (b)

Figure 3.2 Schematic diagram of fabricated flow MRI phantom. (a) represents the phantom with 75% stenosis on U bend tube and 5 mm straight tube; (b) represents the phantom with 50% stenosis on U bend tube and 8 mm straight tube. Both phantoms have dimensions of 240 x 120 x 25 mm^3 size.

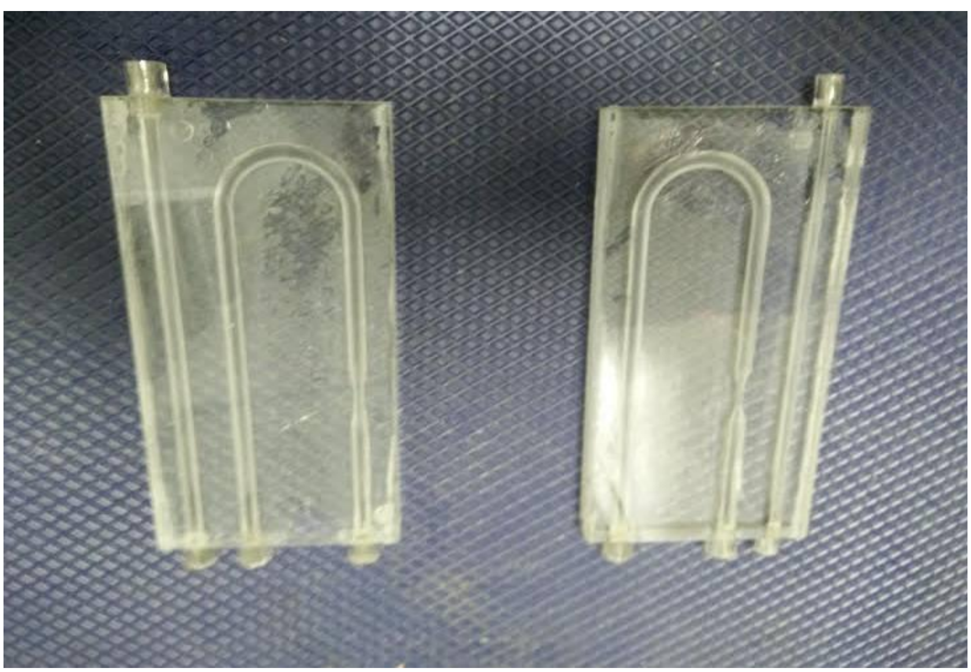


Figure 3.3 The fabricated flow MRI phantoms

The designated phantoms have potential for leakage at every tip of tube connectors. The leakage may damages the MR imaging equipment especially the coil. Therefore, some preventive measure was taken. Firstly, a water catcher basement was introduced to the base of every phantom. The water catcher basement was designed by using polystyrenes and small containers for each tip of tube connectors. Secondly, parts of plastic tubes were permanently secured to every tip tube connectors of the phantom (Figure 3.4).

3.3 Materials

Plastic tube, water pump and container

These materials are connected to the phantom in order to allow the blood mimic mixture circulate through it continuously. Each of the phantoms required four plastic tubes with 640 cm length each to ensure that they reach the water container outside the imaging room. The water pump was placed in the container to pump the blood mimic mixture. The pump was connected to the plastic tubes by the combination of one T-shape connector and two L-shape connectors. A 30 litre container was used as a reservoir to keep blood mimic mixture.

Blood mimic mixtures and agarose gel slab

As addressed by Summers *et al.,* (2005), a 60:40 mixture of distilled water and glycerol has similar viscoscity to the blood (3.5 cP). Therefore, to obtain 25 litre of blood mimic mixture, 15 litre of distilled water were mixed with 10 litre of glycerol. In order to provide background signal, 2% agarose gel slab (200 x 120 x 15 mm^3) was used to sandwich the phantom (Figure 3.5).

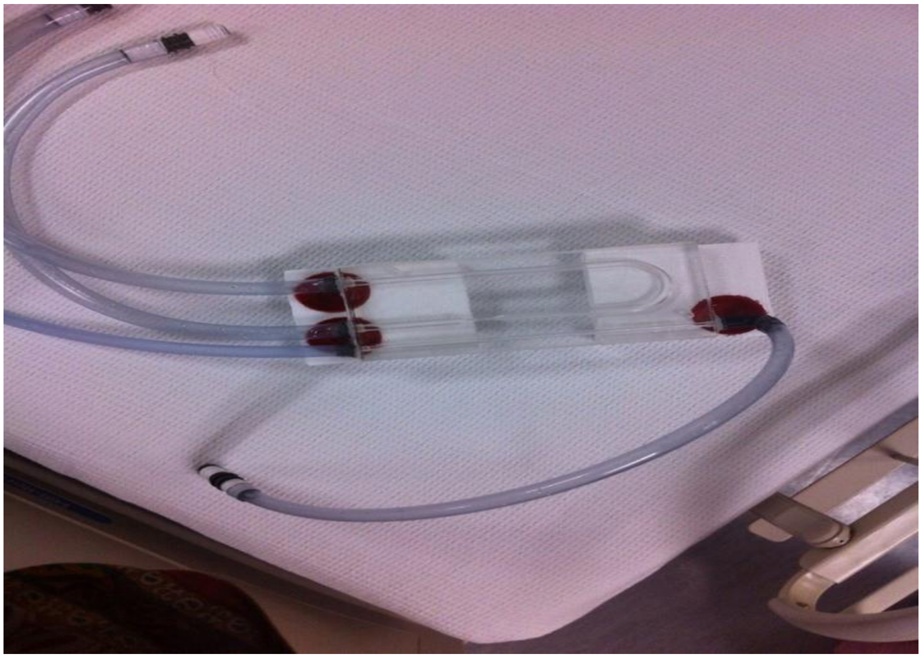


Figure 3.4 The fabricated phantom with the permanent secured of plastic tubes and the water catcher basement

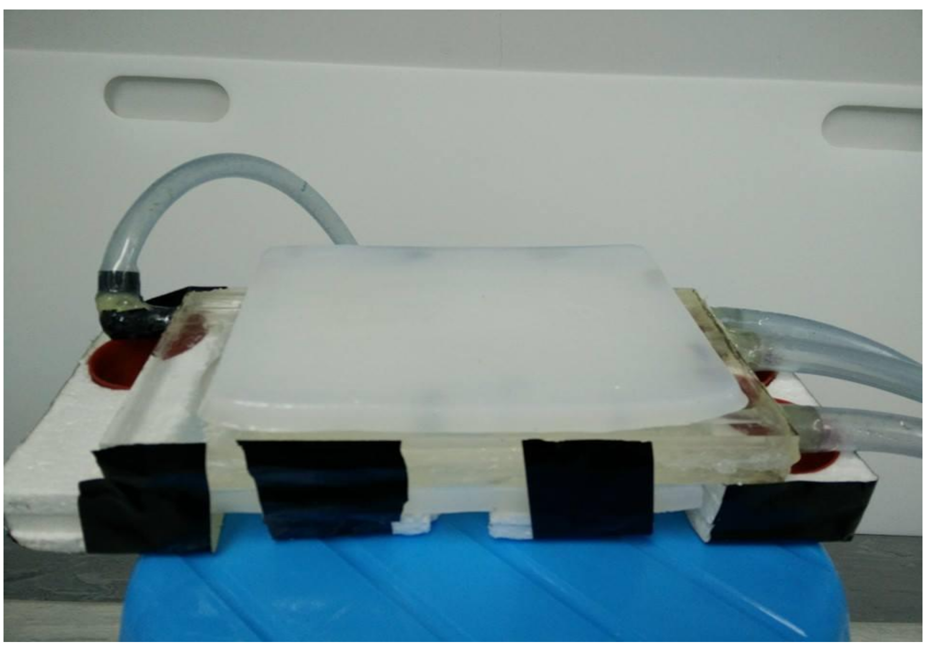


Figure 3.5 The agarose gel slabs placed on the top and bottom of the fabricated flow phantom

3.4 Data collection

3.4.1 Phantom assembly and validation

A mixture of 60:40 distilled water and glycerol that has similar viscosity as blood was prepared in a reservoir outside the imaging room. Water pump was placed in the reservoir. The phantom was placed (one by one at a time) on a 120 cm couch in the imaging room under 16-channel head and neck coil. The phantom was connected to the pump by two inlet plastic tubes. Two outlet plastic tubes from the phantom were left in the reservoir to allow the mixture flow out back into it. The water pump was switched on to circulate the blood mimic mixture with velocity set as 89.7 cm/s for phantom presented with pediatric tube and 76.9 cm/s for phantom with adult tube. The velocity was measured on the straight tubes of each phantom. In order to obtain the accurate measurement of velocity, a dye was used to make the flow of the mixture clearly visualised during the measurement. The time was recorded once the dye reached at the "in" tip of the straight tube till it reached the "out" tip of the tube. The velocity was calculated by dividing the length (25 cm) of the straight tube (tip to tip) with the recorded time. According to Udomphorn *et al.*, (2008), cerebral blood flow velocity of healthy adult is approximately 75 cm/s while in children; the cerebral blood flow velocity is approximately 97 cm/s. The phantoms were validated by ensuring that they were not malfunctions or leaks. The setting was illustrated in Figures 3.6 and 3.7.

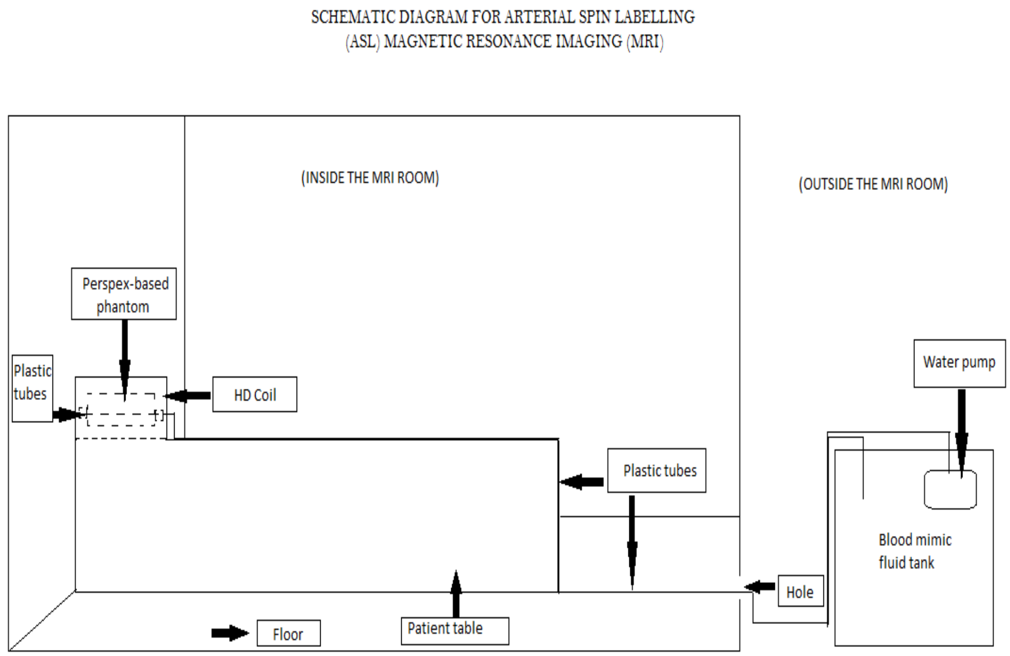


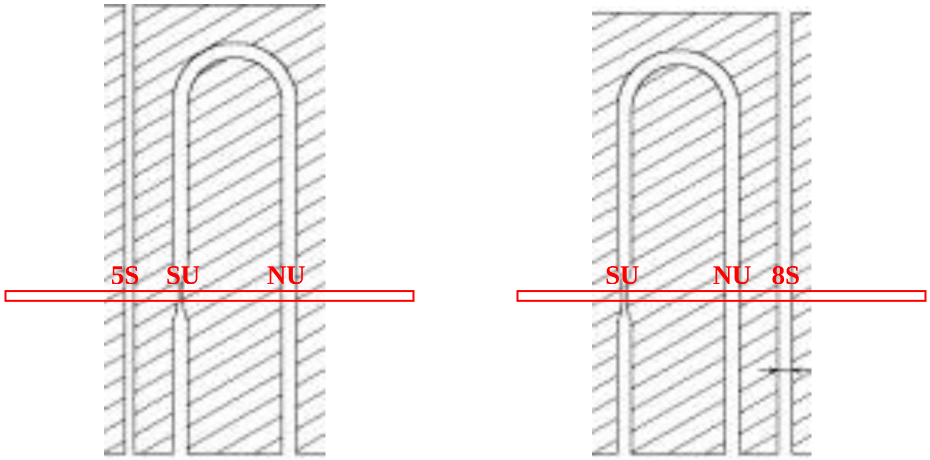
Figure 3.6 Schematic diagram of the phantom set-up in the MR imaging room

Figure 3.7 The experimental phantom set-up in the MR imaging room.

3.4.2 Image acquisition

The image acquisition of the phantoms was performed by using Philips Achieva 3.0 T X-series MR imaging in Radiology Department of Hospital Universiti Sains Malaysia. A flow phantom with 50% stenosis was scanned once the images obtained, the same processes were repeated on the 75% stenosis phantom.

The phantom used 16-channel head and neck coil and scanned using pulsed ASL technique in combination with ASL multiphase and single shot echo planar imaging (EPI) for fast acquisition, sensitivity encoding (SENSE) for fast MR imaging. SENSE was used to reduce echo train length and shorter the scanning time. In the beginning of scanning process, ASL multiphase was used after image localisation to select the label delay (ms) that yield the optimal image visualisation. Within the range of 300 ms to 2200 ms, the optimal image visualisation was obtained at 1645 ms for both phantoms (Figure 3.9). Then, the scanning was proceeded by using standard (single phase) pulsed ASL technique. At label delay 1645 ms, the phantom was scanned by using following parameters in Table 3.1. T1-weighted spin echo (SE) was used to provide anatomical image of the phantoms at position of axial and coronal. Here, the value of FOV and slice thickness will be tested in order to obtain optimal image quality.



(a) (b)

Figure 3.8 Schematic diagram of fabricated flow phantom with the location of PASL labeling, the ROIs. In phantom (a), 5S represents the 5 mm straight tube; SU represents U tube with 75% stenosis; and NU represents U tube without stenosis. In phantom (b), SU represents U tube with 50% stenosis; NU represents U tube without stenosis; and 8S represents the 8 mm straight tube

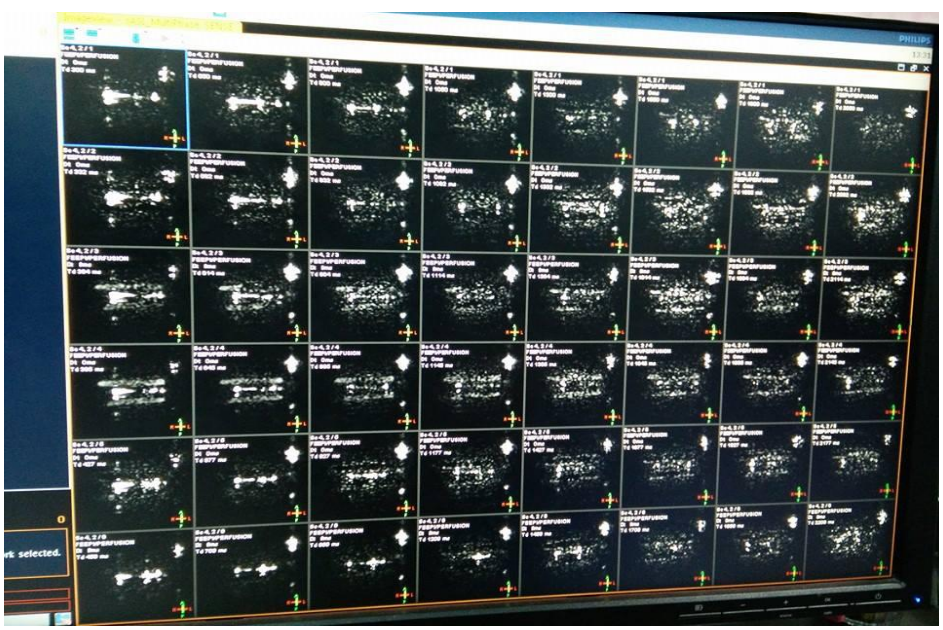


Figure 3.9 The time frame of label delay range from 300 ms to 2200 ms. The label delay that gives optimal image visualisation, 1645 ms was selected as a constant in single phase ASL scan.

Table 3.1 Imaging parameters used for PASL scanning.

Parameters			Values			
Field of view (FOV)	240 x 240 mm^2			320 x 320 mm^2		
Slice thickness	5 mm	7 mm	9 mm	5 mm	7 mm	9 mm
No. of slices	36 slices	28 slices	21 slices	36 slices	28 slices	21 slices
Matrix	68 x 68			88 x 87		
Echo time (TE)	20 ms					
Repetition time (TR)	4000 ms					

3.4.3 Data analysis

Image quality was defined by the value of SNR obtained from the images. The measurement of SNR is obtained by using the equation 1 (Vernikouskaya *et al.*, 2013). The ImageJ software was used to analyses the mean value of ROI, mean value of background and the standard deviation of background.

$$\mathrm{SNR} = \frac{|\bar{I}_{ROI} - \bar{I}_{BG}|}{\sigma_{BG}} \qquad \text{(Eq. 1)}$$

where,

$\bar{I}_{ROI}$ = mean value (signal) of manually drawn region of interest (ROI)

$\bar{I}_{BG}$ = mean value (signal) of background

σ_{BG} = standard deviation of background signal