Progress In Electromagnetics Research M, Vol. 45, 173–183, 2016

On the Feasibility of Assessing Burn Wound Healing without Removal of Dressings Using Radiometric Millimetre-Wave Sensing

Stuart W. Harmer[1, *], Sergiy Shylo[2], Mamta Shah[3], Nicholas J. Bowring[1], and Amani Y. Owda[1]

Abstract—The authors present transmission data, taken at Ka (36 GHz) and W (95 GHz) bands in the millimetre-wave region of the electromagnetic spectrum, for various dressing materials used in the treatment and management of burn wounds. The results show that such materials are highly transparent (typically $> 90\%$ transmission) and, in their dry state, will permit the sensing of the surface of the skin through the thick layers (> 2 cm) of different dressings typically applied in medical treatment of burn wounds. Furthermore, the authors present emissivity data, taken at the same frequency bands, for different regions of human skin on the arm and for samples of chicken flesh with and without skin and before and after localised heat treatment. In vivo human skin has a lower emissivity than chicken flesh samples, 0.3–0.5 compared to 0.6–0.7. However, changes in surface emissivity of chicken samples caused by the short-term application of heat are observable through dressing materials, indicating the feasibility of a millimetre-wave imaging to map changes in tissue emissivity for monitoring the state of burn wounds (and possibly other wounds) non-invasively and without necessitating the removal of the wound dressings.

1. INTRODUCTION

Burns are a very common cause of injury with over a quarter of a million people requiring treatment a year in the UK and costing millions of dollars p.a. [1]. Globally, this figure is far greater. The majority of these burn wounds are partial thickness burns with a potential for spontaneous wound healing from the appendage remnants in the depths of the dermis and surrounding undamaged skin.

The current management of burn wounds requires that the dressings are removed regularly for inspection of the progress of wound healing as well as to detect signs of wound infection. This inevitably causes pain, discomfort and anxiety for the patients, particularly for children. In addition, it also causes distress to the parents/carers and the frequent visits to hospital results in loss of person-hours from work. Furthermore, frequent exposure and handling of the wounds for washing during dressing changes could potentially cause damage to the neo-epithelium covering the wound bed and increase the risk of infection. Although millimetre-wave has been suggested and as a contact probe to assist in the diagnosis of skin cancers [2–8] and there are techniques which utilise infra-red, thermal imagery to assess the wound [9], these require the wound to be exposed. Currently, there are no tools which could assess the state of the healing burn wound without removing the dressings. The materials are layered as follows: the primary dressing, the secondary absorbent layers and the retention layers. A technique that could penetrate dressings and identify the healing status of wounds would be extremely beneficial to both patients and healthcare professionals by reducing the pain, anxiety and distress caused by wound dressing changes as well as reduce healthcare interventional time.

Received 5 November 2015, Accepted 17 December 2015, Scheduled 5 January 2016

* Corresponding author: Stuart William Harmer (swharmer@gmail.com).
[1] School of Engineering, Manchester Metropolitan University, England. [2] Usikov Institute of Radiophysics and Electronics National Academy of Sciences of Ukraine, Kharkiv 61085, Ukraine. [3] NHS, Manchester Department of Plastic Surgery, Royal Manchester Children's Hospital, Oxford Road, Manchester M13 9PT, England.

Several other clinical scenarios would also benefit from such a tool. Patients who have sustained fractures will require plaster casts for prolonged periods of time. Often these patients may also have cutaneous wounds which would require periodic inspection for wound healing status. This inevitably requires the plaster cast to be removed or a window placed in the cast overlying the wound. The mechanics of this interfere with the healing of the underlying fractured bone. Hence, if a system were available to inspect the healing state of the wound through the plaster cast and the dressings, it would be possible to reduce the frequency of plaster cast changes. Another group of patients to benefit would be patients who have relatively clean venous leg ulcers with minimal exudates requiring prolonged compression bandages.

2. MILLIMETRE WAVE RADIOMETRYS

The millimetre-wave (MMW) region of the electromagnetic spectrum lies between 30 GHz and 300 GHz, and perhaps, its best known applications are in security screening [10–13], remote sensing [14–16], automotive radar [17, 18] and communications [19]. When applied to security screening, MMW devices are classified according to whether they transmit artificial MMW radiation, which reflects from the scene being imaged, or, whether they receive only thermally emitted radiation from the scene. The former systems are termed 'active' and the latter 'passive', or radiometric [20]. For both passive and active MMW imagings, it is the electromagnetic properties of textile materials that permit effective imaging of objects concealed under layers of clothing. The complex permittivity of textiles is such that clothing materials (natural and synthetic) are highly transparent [21], allowing differences in the electromagnetic properties of objects (for example, a handgun, knife, explosive material or narcotics) in the MMW band to be observed against the human body. Considerable data are available concerning the phenomenology of millimetre-wave imaging, especially relating to security screening of people [22–24].

Due to their construction, dressing materials used in the treatment of burns and other wounds share similar properties to clothing and because these dressing materials are attached externally to the human body, and the imaging of wounds through dressing materials is a similar problem to security screening. The phenomenology of the wound site is quite different from the phenomenology of most concealed objects of interest, such as metallic weapons, which provide a high contrast in the image due to the large difference in the emissivity and reflectance of metal compared to the human body. The contrast between the undamaged and damaged tissues at a wound site is likely to be smaller and consequently will make imaging of the extent of the wound more difficult than locating a concealed weapon. Although wounds, especially those with infection, may present a higher thermodynamic temperature than healthy tissue, reliance on thermodynamic temperature alone is likely to provide a rather limited tool. For example, a burn, which is healing without infection but is producing excessive scar tissue may not present any thermodynamic temperature contrast. A better approach is to utilise the likely changes in the emissivity between damaged and undamaged tissues. These changes may well result from the altered tissue structure of the wound site. The advantage of this approach over one, which simply maps the thermodynamic temperature, is that the extent of the wound may also be monitored, permitting the healing process to be monitored with recourse to visual inspection. Measurement would be relative, for example, by comparing the brightness temperature of a known area of undamaged skin under the dressing material in close proximity to the wound, and the effects of external factors, such as temperature, humidity and patient perspiration, may be mitigated.

Active MMW imaging is sensitive to alignment and gives an image in which the contrast is due to the differing reflectance in the scene [25]. The sensitivity to alignment is due to the reflectance being dependent on angle of incidence. A small difference in the alignment of a reflecting surface can result in significant differences in the received, reflected radiation and is a common and unwanted phenomenon in active MMW imaging for security screening [20]. Although there is expected to be a difference in the reflectance between undamaged and damaged skins in the MMW band, the sensitivity to alignment may make imaging the wound difficult in practice. Radiometric imaging is better suited for the proposed task as there is no such sensitivity to alignment because skin and dressing materials scatter and radiate nearly isotropically. Radiometric contrast is due to the differences in the brightness temperature (the product of thermodynamic temperature and emissivity) between the wound and undamaged skin.

The receiver output signal, U, is composed of the received power and noise power, which is made up of two components, one from the antenna and the other from the receiver. In the MMW band, the emitted radiation is proportional (with constant of proportionality a) to the product of the thermodynamic temperature of the emitter and the emissivity of the surface, and this quantity is the 'brightness temperature' T' and is given by,

$$U = a\left(T' + T_N\right) \tag{1}$$

The noise temperature is given by

$$T_N = \left[\frac{1}{\eta_a} - 1\right] T_a + \frac{(NF-1)}{\eta_a} T_0 \tag{2}$$

where, η_a is the antenna efficiency, T_a the temperature of the antenna, NF the noise figure of the receiver and T_0 the ambient temperature. The brightness temperature of the sample, T', is dependent upon the electromagnetic properties of the materials that comprise the sample and their thermodynamic temperatures. The dressed wound may be approximately modelled with a planar three-layer system, as shown in Figure 1. The brightness temperature of the sample is determined by the external radiation, which is reflected from the sample; the radiation which is transmitted from the skin, through the dressing, and subsequently received and the radiation which is emitted from the dressing layer itself and then received. In the ideal case ($\eta_a = 1$ and $NF = 1$) the receiver noise temperature is equal to that of the surroundings, T_0, this receiver noise radiation is reflected from the sample and contributes to the received signal (see [26] for further information).

$$T' = T_0 R(1-f) + T_N R f + T_D A + T_S \varepsilon \tag{3}$$

where, T_D and T_S are the temperatures of the dressing layer and wound/skin, respectively. R is the reflectance of the dressed wound, A the absorptance (emissivity) of the dressing layer and ε the emissivity of the wound/skin. The parameter f is the fraction of the total radiation blocked from being reflected from the sample by the antenna, and the value of f depends upon the antenna and its proximity to the sample. Conservation of energy for the system is expressed as,

$$1 = R + A + \varepsilon \tag{4}$$

Using Eq. (3) for an ideal receiver, with $T_N = T_0$ (which may be realised by placing a waveguide circulator between the antenna and receiver) and for very low loss dressing materials, $A \approx 0$, Eq. (3) is simplified to,

$$T' \approx T_0(1-\varepsilon) + T_S \varepsilon \tag{5}$$

The receiver output (Eq. (1)) can be written in terms of the 3-layer system's brightness temperature (Eq. (3)) as,

$$U \approx a\left(T_0(1-\varepsilon) + T_S\varepsilon + T_N\right) \tag{6}$$

Determination of the emissivity is realised by means of a two point (linear) calibration with 'hot' and 'cold' black bodies; one at ambient temperature, T_0 and the other held at a higher temperature, T_H. From Eqs. (1) and (6), these measurements give,

$$U_H = a\left(T_H + T_N\right) \tag{7}$$
$$U_C = a\left(T_0 + T_N\right) \tag{8}$$

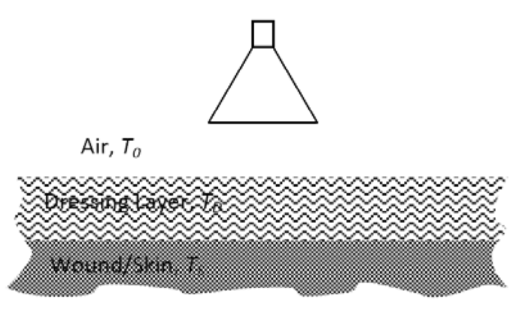

Figure 1. The model of the dressed wound, comprising three layers (air, dressing and the wound/skin).

Subtraction of Eqs. (7) and (8) provides the constant of proportionality, a, as,

$$a = \frac{U_H - U_C}{T_H - T_0} \tag{9}$$

The value of aT_N may be determined directly from either Eq. (7) or Eq. (8) as,

$$aT_N = U_H - aT_H \tag{10}$$

So that the emissivity of an unknown sample may be determined using Eqs. (6), (7) and (10) as,

$$\varepsilon \approx \frac{(U - U_C)(T_H - T_0)}{(U_H - U_C)(T_S - T_0)} \tag{11}$$

The thermodynamic temperature of the sample, T_S, is required in Eq. (11) to compute the emissivity and is measured using suitable probes. Since the temperature may well vary over the wound site, a thermodynamic temperature map, $T_S(x, y)$, is required to obtain an emissivity map $\varepsilon(x, y)$. However, with current dressings this is not feasible. Because the images have an absolute brightness temperature scale, due to the two-point calibration process, the calculation of emissivity (by measurement of the skin's thermodynamic temperature) is unnecessary in a practical system, and the brightness temperature will show relative variations in the surface of the wound. Furthermore, multiple images, taken different times, will provide information on the wound healing process by measuring the change in size and shape of the wound and may be able to reveal the incidence of infection or other disease process.

The largest error is likely to be that of measuring the temperature of the skin, and the error due to noise can be reduced by increasing the integration time during which the sensor receives radiation from the area under observation [20]. The relative error in emissivity, $\Delta\varepsilon$, associated with measurement error of the temperature of the skin, ΔT_S, is,

$$\left|\frac{\Delta\varepsilon}{\varepsilon}\right| \approx \left|\frac{\Delta T_S}{T_0 - T_S}\right| \tag{12}$$

Therefore, a low contrast between ambient and skin temperature will increase the error in the emissivity measurement, and uncertainty in skin/wound temperature will translate into uncertainty in the permittivity. This is likely to be the limiting factor in determining emissivity since it may not be practicable to accurately measure the temperature of the surface of the skin/wound under the dressing at or lower than the spatial resolution of the imaging system. With an error of 1 K in skin temperature and a difference between background and skin temperatures of ~ 10 K the relative error in permittivity is ~ 0.05.

3. EXPERIMENTAL DESCRIPTION

The horn antenna may be located in three regions: 'hot' black body, 'cold' black body or SUT, see Figure 2. The first two regions are used to provide a calibration, as outlined in Section 2. The third region contains the SUT, and the output of the receiver may be used, in conjunction with the calibration data, to obtain the emissivity or loss of the SUT. The horn antenna is fixed in the laboratory, and the other equipment is placed upon a movable cart, which travels on rails, thereby permitting the system to be calibrated prior to examining the SUT. The receiver output signal is registered by digital data acquisition. A W-band waveguide circulator is placed between horn antenna and receiver input, so that noise radiation emitted from the antenna to the SUT is the same in intensity as background radiation with air temperature T_0. Dressing materials are placed directly onto the SUT when required. An electric heating element is used to heat the SUT to a desired bulk temperature, and temperature probes are used to measure the bulk and surface temperatures of the SUT. This information along with the calibration allows the emissivity of the SUT to be calculated using Eq. (11). The parameters for the experimental apparatus are given in Table 1.

4. RESULTS & DISCUSSION

Although it is very well known that MMW radiation is minimally attenuated in textiles used in clothing, with the exception of leather [21] and [27], investigation of the transparency of medical dressing materials

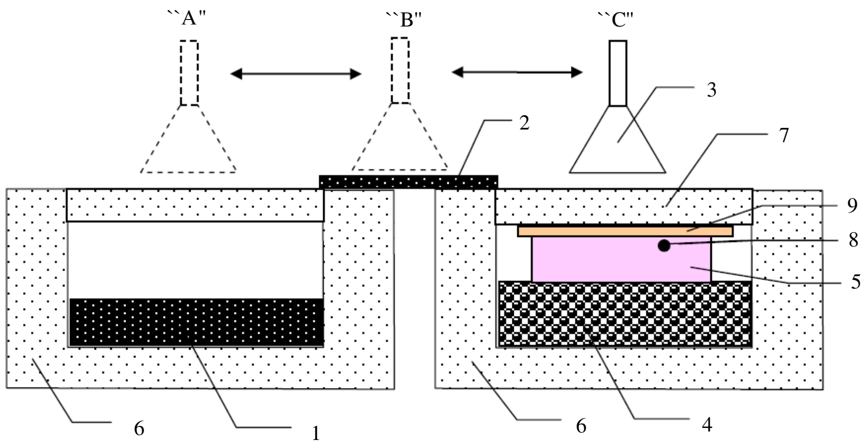

Figure 2. The main elements are: 1 — 'Hot' black body emitter (carbon loaded foam) at a temperature $\sim$ 54C; 2 — 'Cold' black body emitter in thermodynamic equilibrium with air temperature $\sim$ 20C; 3 — Horn antenna of the measuring radiometer; this can be located in 3 regions — A, or B, or C; 4 — Electric heating element; 5 — Sample Under Test (SUT); 6 — Insulating chamber $20 \times 20 \times 10$ cm (polystyrene foam); 7 — Radio transparent window (polystyrene foam); 8 — sensor for mean temperature of a sample tissue indication; 9 — Dressing materials placed over the SUT (when required).

Table 1. System Parameters.

Parameter	Value	Remarks
Radiometric system		
Centre frequency (f_0)	95 GHz	
High frequency band	±1 GHz	relative to (f_0)
Temperature sensitivity	0.2 K	
Integration time	1 s	
Aperture size of the receiving horn antenna ($E \times H$), mm^2	$15 \times 20\,\text{mm}^2$	
Hot Load		
Reflection coefficient	−20 dB	at W band
Emissivity	0.99	at W band
Viewing angles	±60°	relative to the normal to the plane of the radiotransparent window
Average operating temperature	+54°C (327 K)	corresponds to brightness of output radiation
Temperature variation	±1.5 K	

used in the treatment of burn injuries was undertaken, as there is little available information for these materials in the literature. Measurements to determine the emissivity, or loss, for the dressing samples listed in Table 2 were made at W-band (95 GHz centre frequency) and Ka-band (36.6 GHz centre frequency), and these measurements are based on Eq. (11). The skin is now replaced by the hot calibration load (i.e., the dressing material is placed directly onto the hot calibration load), $T_S = T_H$, so that, for dressing materials alone,

$$\varepsilon \approx \frac{(U - U_C)}{(U_H - U_C)} \tag{13}$$

Here, the ambient temperature was $\sim$ 25C, and the hot calibration load had a temperature of $\sim$ 54C.

It may be seen that the losses at Ka and W bands are not dissimilar; this could be due to the large

Table 2. Transmission of MMW, at Ka and W bands, through dressings used in the treatment of burn injuries.

Dressing Type	Sample number	Transmission %		Loss (dB)		Comments
		Ka-band	W-band	Ka-band	W-band	
Gauze	1	96.4	94.0	0.16	0.26	All dressings are dry and removed from protective packaging prior to measurement. The air temperature was ∼ 25C during measurements. 6 layers of gauze are used in the measurements
Light Support Bandage Type 2 (BP cotton stretch, 10 cm × 4.5 m)	2	95.2	93.8	0.21	0.28	1 layer used
Light Support Bandage Type 2 (BP cotton stretch, 10 cm × 4.5 m)	3	91.9	88.9	0.36	0.51	2 layers used
Light Support Bandage Type 2 (BP cotton stretch, 10 cm × 4.5 m)	4	86.6	82.5	0.62	0.84	3 layers used
Elasticated stocking bandage	5	94.7	88.6	0.23	0.82	2 layers used (in practice, only one layer will be present in patients)
Non-adherent Clear Wound Dressing (Tefla™)	6	100	100	0	0	1 layer used
Temporary Wound Dressing (Biobrane®)	7	100	100	0	0	1 layer used

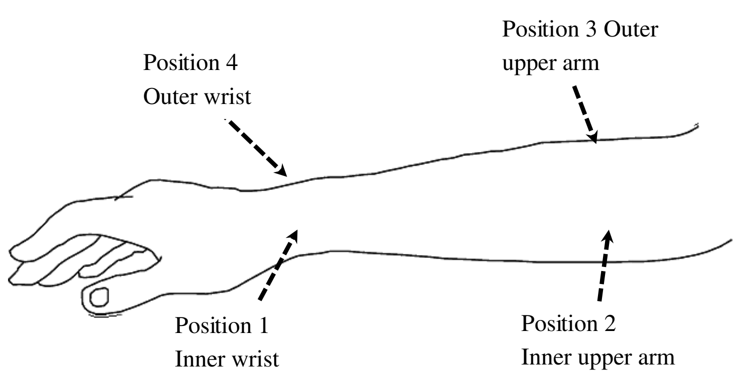

Figure 3. Four positions on the lower arm where emissivity of human skin at W-band was measured on eight uninjured volunteers. Each position was measured several times and the mean and standard deviation of the results are given presented in Figure 4.

scale porosity of the samples, especially samples 1–4, which may cause more scattering at Ka band (where structure size is ∼ wavelength) than at W-band (where structure size is > wavelength). Sample 5 (without the stretching of fabric) has a smaller scale structure and therefore results in more loss at

W band. Ka-band measurements of the emissivity of chicken tissue (breast) heated to a temperature of ~ 37C with give values of 0.31 and 0.36 for undressed and dressed samples. The dressings applied were four layers of burns gauze and one layer of Light Support Bandage Type 2. There is some increase in emissivity associated with the application of dressing materials, but this increase is within the error of the measurement.

To assess the feasibility of measuring changes in emissivity at W-band, eight uninjured volunteers were measured at four points on their lower arm, see Figure 3. This will allow an assessment of the expected range of permittivity values of the human skin on the arm and the likely variation from person to person.

In the uninjured skin there is variation in emissivity from location to location of the lower arm, and this variation is likely to be similar across other parts of the body. The variation is quite large for persons 5, 6 and 7, probably due to greater presence of subcutaneous fat in these people. The variation (standard deviation) calculated by taking multiple measurements is 0.027 for person one and 0.082 for person seven. Variations in emissivity are important because the emissivity map will change between points on the skin to a greater or lesser degree. These changes may prevent useful imagery of wounds if the variations are larger or commensurate to the changes in emissivity which result from the wound. However, with a burn or other wound the change in emissivity between uninjured and damaged tissue is expected to be a well-defined interface taking place over distances of ~ 1 cm, whereas changes due to variations in composition of healthy tissue are expected to be less localised, taking place over distances of ~ 10 cm.

Assessing the likely changes in emissivity of human skin which has suffered from burn injury requires patients who have burn injuries. A simpler, though less specific, experiment was to select a biological tissue from a dead animal and use this as a 'phantom' for living human tissue. Chicken (breast and leg tissue, with and without skin) was selected for this purpose. The chicken samples were raised to human body temperature ~ 37C and their emissivities measured at W band in the MMW spectrum. The dressing materials, four layers of burns gauze and one layer of Light Support Bandage Type 2 (BP cotton stretch), were applied, and their effect on the emissivity is seen to be small in both Ka and W bands (see Figure 5).

Multiple measurements were taken for each case (A, B, C and D), and the mean and standard deviation of these are presented Figure 5. Firstly, the emissivity of the chicken samples is about two times larger at W-band (~ 0.6) than at Ka-band (~ 0.3), so W-band measurement will likely be more sensitive and offer a higher spatial resolution in imaging due to the shorter wavelength. The additional losses in dressing materials at W-band are not so great as to reduce contrast, so W-band would seem to be preferable to Ka band in all respects except, perhaps, the higher cost of components required to build imaging systems.

The effects of skin on the tissue samples are observable, comparing A and B in Figure 5. The presence of skin reduces the emissivity which is consistent with the presence of subcutaneous fat having

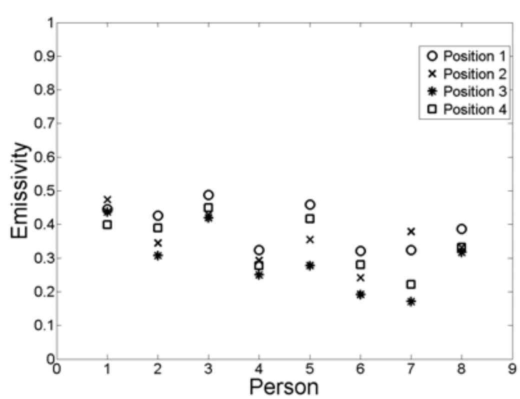

Figure 4. Mean values of the measured emissivity for of eight people at four different points on their lower arms. The people's gender and age is as follows: 1) M, 62; 2) M, 23; 3) M, 61; 4) M, 65; 5) M, 67; 6) F, 55; 7) F, 58; 8) M, 61.

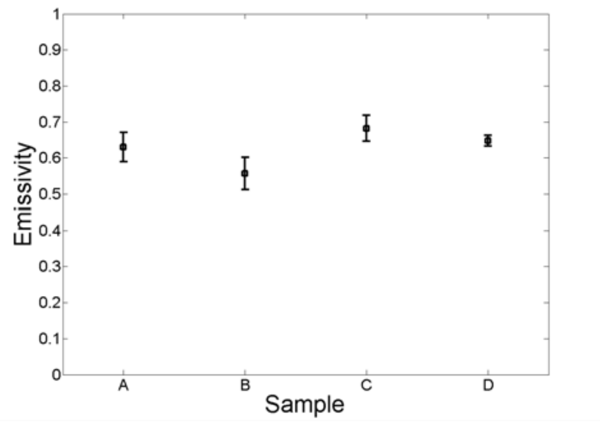

Figure 5. Measured values of emissivity and standard deviation bars for chicken samples at ∼ 37C. The samples are: A, tissue without skin and without dressing materials; B, tissue with skin and without dressing materials; C, tissue without skin but with dressing materials and D, tissue with skin and dressing materials.

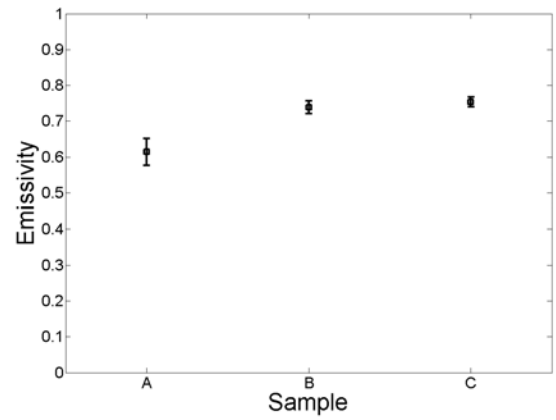

Figure 6. Samples of chicken tissue with skin which have undergone high temperature heating with air at 200C. Sample A, control without heating; B, heat treatment applied for 60 s; C, heat treatment applied to sample twice. The samples are measured without dressing materials. The standard deviation calculated from multiple measurements is displayed as error bars.

Table 3. Electromagnetic measurements of human skin in the millimetre wave band for medical diagnosis.

Type of the Skin	Measured Quantity	Reference
Skin Cancer	Complex Permittivity at 20–70 GHz	[28]
Healthy Skin and Burn Wounds	Complex permittivity at 30–40 GHz	[29]
Healthy Skin	Complex Permittivity and Emissivity at 20–36 GHz	[30]
Dry Skin	Complex Permittivity at 57–76 GHz	[23]

a higher reflectance than tissue without fat. The effects of dressings (see C and D in Figure 5) slightly increase the emissivity, due to increased losses and increased transmission from the skin to air by virtue of reducing the discontinuity in impedance at the skin/air interface. However, the lower emissivity of the sample with skin is still seen through the dressing materials. The chicken tissue samples have a significantly higher emissivity than in-vivo human skin (about 50% larger) and therefore is not an accurate phantom. Furthermore, the response of living tissue to burn injuries is different from that of the non-living samples. However, alteration of the sample emissivity by application of localised heat demonstrated that changes in the surface and near surface layers of biological tissue were measurable through wound dressing materials.

A chicken tissue sample with skin was exposed to hot air (200C) for 60 s to alter the surface material by denaturing proteins in the tissue, and the sample was then allowed to cool and maintained at body temperature by electrical heating. The emissivity was measured, without dressing materials, before exposure to heat, after one exposure to heat and then again after a second exposure to heat, see Figure 6. Application of heat increases the emissivity of the sample from 0.593 to 0.724, and a second application leaves the emissivity almost unchanged.

The application of millimetre wave to medical screening for burns and skin cancer is summarised in Table 3. In most articles in this field, the complex permittivity is the measured quantity; for passive imaging, the emissivity is more useful. Significant differences in complex permittivity between healthy and diseased or damaged skin do not necessarily imply commensurate differences in emissivity.

5. CONCLUSION

Dressings used in the medical management of burn injuries were measured for losses at two commonly used bands in the MMW spectrum, Ka and W-bands. Both bands have low attenuation in all dressing materials tested, in the unused state, allowing for the possibility that the healing process of such wounds may be monitored without recourse to removing the dressing, as current medical practice.

The emissivity of samples of chicken, with and without skin, was measured to ascertain whether these would make mimic living human tissue. Although the chicken samples have a higher emissivity than living human skin and tissue, the capability to measure the effect of localised alteration to the emissivity of chicken samples was observed. Chicken samples were heated with hot air for a short period of time, and the emissivity of the samples was measured before and after such treatment. Heat treatment produces an increase in the emissivity of chicken samples of $\sim 25\%$ by altering the chemistry and structure of the surface and near surface tissue, and this change is observable through the dressing materials used in the management of burn injuries, allowing the size of the 'damaged' area to be observed.

Measurements of the variation in emissivity of human skin on the lower arm show significant variation from person to person and, more importantly, significant variation at different positions on the arm of individuals. These variations in emissivity are similar in magnitude to those produced by the localised high temperature heating of the chicken sample. However, these natural variations occur over longer length scales than would be the case for a localised wound site, so the edges of wounds are likely to be observable through the typical dressing materials applied to burn injuries. Changes in ambient conditions such as temperature and humidity and variations in the skin due to perspiration or surface temperatures will alter the mean brightness temperature of the scene but the contrast between damaged and undamaged skin will likely be preserved. In this way, a time series of images taken on different days will present different mean brightness temperatures, but the edges and features of the wound should remain observable due to relative differences in brightness temperature. The most likely serious limitation will be the deposition of biological matter from the wound into the dressing materials. These exudates may prevent effective measurement of the wound's emissivity, although there may be a correlation between exudate depositions and the state of the wound, and so the technique could still provide useful information on the state of the wound without necessitating removal of the entire dressing.

REFERENCES

1. The Child Accident Prevention Trust (CAPT), "The costs of burns," *Working Together for Safer Children*, Aug. 2013, Available: http://www.makingthelink.net/tools/costs-child-accidents/costs-burns, Accessed: Sep. 12, 2015.
2. Arbab, M. H., T. C. Dickey, D. P. Winebrenner, A. Chen, M. B. Klein, and P. D. Mourad, "Terahertz reflectometry of burn wounds in a rat model," *Biomedical Optics Express*, Vol. 2, No. 8, 2339–2347, Aug. 2011.
3. Smulders, P. F. M., "Analysis of human skin tissue by millimeter-wave reflectometry," *Skin Research and Technology*, Vol. 19, No. 1, 209–216, Apr. 2012, DOI: 10.1111/j.1600-0846.2012.00629.x.
4. Kharkovsky, S., M. T. Ghasr, M. A. Abou-Khousa, and R. Zoughi, "Near-field microwave and mm-wave noninvasive diagnosis of human skin," *International Workshop on Medical Measurements and Applications*, 5–7, Italy, May 2009.
5. Siegel, P. H., "Terahertz technology in biology and medicine," *IEEE Transactions on Microwave Theory and Techniques*, Vol. 52, No. 10, 2438–2447, Oct. 2004.
6. Essen, H., J. M. Essen, D. Nuessler, A. Hommes, C. Krebs, N. Fatihi, and T. Buzug, "Monitoring of wound healing by millimetre wave imaging," *35th International Conference on Infrared, Millimeter, and Terahertz Waves*, 1–2, Sep. 2010.
7. Siegel, P. H., "Microwave symposium digest," *IEEE MTT-S International*, Vol. 3, 1575–1578, Jun. 2004, ISSN: 0149-645X, Print ISBN: 0-7803-8331-1 INSPEC, Accession Number: 8058048, DOI:10.1109/MWSYM.2004.1338880.

8. Bardati, F., V. J. Brown, M. P. Ross, and P. Tognolatti, "Microwave radiometry for medical thermal imaging: Theory and experiment," *IEEE MTT-S International in Microwave Symposium Digest*, Vol. 3, 1287–1290, Jun. 1992, ISSN: 0149-645X, Print ISBN: 0-7803-0611-2 INSPEC, Accession Number: 4296960, DOI: 10.1109/MWSYM.
9. Dziewonski, M., "Planimetry of thermograms in diagnosis of burn wounds," *Scientific Research of the Institute of Mathematics and Computer Science*, Vol. 8, No. 1, 33–38, 2009.
10. Federici, J. F., B. Schulkin, F. Huang, D. Gary, R. Barat, F. Oliveira, and D. Zimdars, "THz imaging and sensing for security applications — Explosives, weapons and drugs," *Semiconductor Science and Technology*, Vol. 20, No. 7, S266–S280, Jun. 2005.
11. Appleby, R. and R. N. Anderton, "Millimeter-wave and submillimeter-wave imaging for security and surveillance," *IEEE Publisher*, Vol. 95, No. 8, 1683–1690, Aug. 2007, ISSN: 0018-9219 INSPEC, Accession Number: 9633436, DOI:10.1109/JPROC.2007.898832.
12. Sheen, D. M., D. L. McMakin, and T. E. Hall, "Three-dimensional millimeter-wave imaging for concealed weapon detection," *IEEE Transactions on Microwave Theory and Techniques*, Vol. 49, No. 9, 1581–1592, Sep. 2001.
13. Oka, S., H. Togo, N. Kukutsu, and T. Nagatsuma, "Latest trends in millimeter-wave imaging technology," *Progress In Electromagnetics Research Letters*, Vol. 1, 197–204, 2008.
14. Nashashibi, A. Y., K. Sarabandi, P. Frantzis, R. D. De Roo, and F. T. Ulaby, "An ultrafast wide-band millimeter-wave (mmw) polarimetric radar for remote sensing applications," *IEEE Transactions on Geoscience and Remote Sensing*, Vol. 40, No. 8, 1777–1786, Aug. 2002, ISSN: 0196-2892 INSPEC, Accession Number: 7416289, DOI:10.1109/TGRS.2002.802462.
15. Appleby, R., R. N. Anderton, N. H. Thomson, and J. W. Jack, "The design of a real-time 94-GHz passive millimetre-wave imager for helicopter operations," *Passive Millimetre-wave and Terahertz Imaging and Technology*, Vol. 5619, 38–46, Dec. 2004, DOI:10.1117/12.581336, Available: http://dx.doi.org/10.1117/12.581336, Accessed: Sep. 13, 2015.
16. Coward, P. and R. Appleby, "Comparison of passive millimeter-wave and IR imagery in a nautical environment," *Passive Millimeter-wave Imaging Technology XII*, Vol. 7309, No. 730904, Apr. 2009, DOI:10.1117/12.819852, Available: http://dx.doi.org/10.1117/12.819852, Accessed: Sep. 13, 2015.
17. Hasch, J., E. Topak, R. Schnabel, T. Zwick, R. Weigel, and C. Waldschmid, "Millimeter-wave technology for automotive radar sensors in the 77 GHz frequency band," *IEEE Transactions on Microwave Theory and Techniques*, Vol. 60, No. 3, 845–860, Mar. 2012, ISSN: 0018-9480 INSPEC, Accession Number: 12571080, DOI:10.1109/TMTT.2011.2178427.
18. Wengerl, J., "Automotive mm-wave radar: Status and trends in system design and technology," *Automotive Radar and Navigation Techniques, IEE Colloquium. IET*, Ref. No. 1998/230, 1–7, 1998, INSPEC, Accession Number: 5877493, DOI:10.1049/ic:19980188.
19. Pi, Z. and F. Khan, "An introduction to millimeter-wave mobile broadband systems," *IEEE Communications Magazine*, Vol. 49, No. 6, 101–107, Jun. 2011, ISSN: 0163-6804 INSPEC, Accession Number: 12036181, DOI:10.1109/MCOM.2011.5783993.
20. Kemp, M. C., "Millimetre wave and terahertz technology for the detection of concealed threats: A review," *Optics and Photonics for Counterterrorism and Crime Fighting II*, Vol. 6402, Sep. 2006, DOI:10.1117/12.692612, Available: http://dx.doi.org/10.1117/12.692612, Accessed: Sep. 14, 2015.
21. Lamb, J. W., "Miscellaneous data on materials for millimetre and submillimetre optics," *International Journal of Infrared and Millimeter Waves*, Vol. 17, No. 12, 1997–2034, Dec. 1996.
22. Alekseev, S. I. and M. C. Ziskin, "Human skin permittivity determined by millimeter wave reflection measurements," *Bioelectromagnetics*, Vol. 28, No. 5, 331–339, Jul. 2007, DOI: 10.1002/bem.20308, Available: http://onlinelibrary.wiley.com/, Accessed: Sep. 14, 2015.
23. Alabaster, C. M., "The microwave properties of tissue and other lossy dielectrics," PhD Thesis, Cranfield University, Mar. 2004.
24. Zastrow, E., S. K. Davis, M. Lazebnik, F. Kelcz, B. D. Van Veen, and S. C. Hagness, "Development of anatomically realistic numerical breast phantoms with accurate dielectric properties for modeling microwave interactions with the human breast," *IEEE Transactions on Biomedical Engineering*, Vol. 55, No. 12, 2792-2800, Dec. 2008.

25. Harmer, S. W., N. Bowring, D. Andrews, N. D. Rezgui, M. Southgate, and S. Smith, "A review of nonimaging stand-off concealed threat detection with millimeter-wave radar," *IEEE Microwave Magazine*, Vol. 13, No. 1, 160–167, Feb. 2012, ISSN: 1527-3342 INSPEC, Accession Number: 12479283, DOI:10.1109/MMM.2011.2174125.
26. Bardati, F. and D. Solimini, "Radiometric sensing of biological layered media," *Radio Science*, Vol. 18, No. 6, 1393–1401, Dec. 1983, DOI: 10.1029/RS018i006p01393.
27. Harmer, S. W., N. Rezgui, N. Bowring, Z. Luklinska, and G. Ren, "Determination of the complex permittivity of textiles and leather in the 14–40 GHz millimetre-wave band using a free-wave transmittance only method," *IET Microwaves, Antennas & Propagation*, Vol. 2, No. 6, 606–614, Sep. 2008, ISSN: 1751-8725 INSPEC, Accession Number: 10158942, DOI:10.1049/iet-map:20070235.
28. Aminzadeh, R., M. Saviz, and A. A. Shishegar, "Dielectric properties estimation of normal and malignant skin tissues ats millimeter-wave frequencies using effective medium theory," *22nd Iranian Conference on Electrical Engineering (ICEE), 2014*, 1657–1661, IEEE, May 2014.
29. Olga, B. L., Y. Nikawa, W. Snyder, J. Lin, and K. Mizuno, "Novel microwave and millimeter-wave biomedical applications," *4th International Conference on Telecommunications in Modern Satellite, Cable and Broadcasting Services, 1999*, Vol. 1, 186–193, IEEE, Oct. 1999.
30. Tamyis, N. M., D. K. Ghodgaonkar, M. N. Taib, and W. T. Wui, "Dielectric properties of human skin in vivo in the frequency range 20–38 GHz for 42 healthy volunteers," *Proc. of the 28th URSI General Assembly*, 2005.